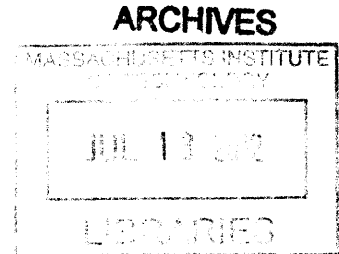


Effects of a Powered Ankle Prosthesis on Shock Absorption and Residual Limb/Socket Interface Pressure

By

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B.S. Physics
Morehouse College, 2010



Submitted to the Program in Media Arts and Sciences, School of Architecture and Planning,
in partial fulfillment of the requirements for the degree of

Master of Science in Media Arts and Sciences

at the

MASSACHUSETTS INSTITUTE OF TECHNOLOGY

June 2012

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Abstract

Lower-extremity amputees face potentially serious post-operative complications, including increased risk of further amputations, excessive stress on both limbs, and discomfort at the stump/socket interface. State of the art, passive prostheses have improved many negative consequences associated with lower-limb loss, but we believe the limit of uninformed elastic prostheses has been reached.

Further strides require a more biomimetic approach. Through integration of “smart” technology (sensors and actuators), a new phase of bionic lower-limb prostheses is upon us, which enables prosthetic devices to more closely mimic biological behavior by generating human-like responses and power outputs. The closer we come to natural biology, gait abnormalities in amputees will decline.

This project compares the first bionic ankle prosthesis to commonly used passive prostheses to determine how more biomimetic adaptability and work generation in the prosthetic joint affects discomfort and joint stress. We have put forth several metrics to describe discomfort (elements of shock absorption, pressure distribution, etc.) and will conduct level-ground walking tests with three unilateral amputee subjects using both passive and power devices. We hope to make a case for the pursuit of more biomimetic designs for rehabilitative devices, by showing a positive effect on “comfort” and a restoration of normal gait dynamics when using a bionic ankle prosthesis.

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List of Terms

Biomechanics – The study of the structure and function of biological systems by means of the methods of mechanics.

Gait Analysis – The study of walking or other types of ambulation.

Energetics – A branch of mechanics that deals primarily with energy and its transformations.

Lower Extremity Amputee (LEA) – A person with an amputation of the lower limb.

Transtibial Amputation – A lower extremity amputation occurring at the tibia (between the ankle and the knee). Also called Below-Knee Amputation.

Unilateral – Relating to, occurring on, or affecting only one side of an organ or structure of the body.

Biomimetic – The study of the structure and function of biological systems as models for design and engineering.

Socket – The part of a prosthesis into which the residual limb fits. It forms the connection between the person and the prosthesis.

Sagittal Plane – Vertical plane of the body that divides it into right and left halves.

Flexion – A decrease in joint angle. The opposite of extension.

Extension – An increase in joint angle, usually resulting in straightening of the limb. It is the opposite of flexion.

Adduction – Inward motion that draws the limb toward the center axis (sagittal plane) of the body. It is the opposite of abduction.

Abduction – Opposite of adduction. Outward motion of the limb away from the center axis (sagittal plane) of the body.

Chapter 1

Introduction

Over 100,000 lower extremity amputations occur every year in the United States alone [Uustal 2006]. With major causes of amputations – war, vascular disease, and natural disasters – prevalent both domestically and abroad, the need for effective prostheses will continue to be a huge issue for years to come. Great strides have been made since the clunky wooden attachments of old, but much room exists for further enhancements in the field. Early innovation in prosthetic designs aimed to be biomimetic in appearance, but offered little beyond this. Recent innovation attempts to be more biomimetic in function. This new approach, incorporating advanced technology to enhance functionality, has taken the field past the days of cumbersome space fillers and into a day of “smart” devices with more versatile capabilities. However, over time as prostheses for the lower extremities have become more refined and sophisticated, some common complications persist.

Human gait is a highly efficient process sometimes taken for granted by the able-bodied. Those with intact biological limbs and no muscle deficiencies are unaware of the complex biophysics involved. Muscles must fire in harmony with perfect timing [Perry 1992]. Joints must act as both torque sources and energy dissipators. Subtle changes in joint or muscle behavior and seemingly small perturbations can cause significant differences in gait patterns, making locomotion far less efficient. Therefore, a significant abnormality, such as an amputation, creates a substantial reduction in legged locomotion efficiency.

Current state-of-the-art prosthesis technology has brought amputee gait closer to normal efficiency levels than past devices. However, lower extremity amputees (LEA) continue to experience gait pathologies, with deficiencies in rhythm, timing, kinetics, and energetics. Gait pathologies along with other functions lost with amputation put LEA at an increased risk of several medical complications; including further amputation, skin disorders, and most commonly excessive pain/wear in both legs [Perry 1992, Levy 1980]. Eventually, these complications lead to serious joint degradations, often times spreading beyond the lower extremities and into the back [Perry 1992]. Faced with these issues, many amputees decide to greatly reduce or even completely discontinue use of the prosthesis, leaving them essentially immobile.

The fundamental goal of all prostheses is to restore natural function to amputees. So far, none are successful. It is important, though, to track their progression, measuring how closely they come to achieving the ultimate objective. The new wave of “smart” prostheses seems to be a step closer to realizing the function of the human body and biological joints. In reference to prostheses for LEA, the hope is that new devices will reduce gait pathologies and improve gait efficiency, alleviating short- and long-term medical risks. To analyze this, one can put together experiments to test new devices against older ones, to see how a “smarter” and more biomimetic prosthesis affects amputee gait. Hopefully, by evaluating quantitative gait measurements, insight can be gained as to ways to improve prostheses to reduce the risks associated with them.

1.1 Thesis Objectives

The goal of this thesis is to uncover how an ankle prosthesis that more closely mimics the behavior of the biological joint affects the distribution of pressure and applied stresses on both lower extremities. We believe that pressure and stress are related to long-term joint degradation and short-term discomfort, so throughout this thesis we will use the term “comfort” as a blanket term to describe applied pressure and stress along the legs.

This study will analyze how a more functionally biomimetic prosthesis affects gait patterns, evokes a more natural feeling gait, and ultimately serves comfort for unilateral transtibial amputees walking on level terrain. To sufficiently measure our findings we must first put forth metrics to define and quantify “comfort”. The metrics to be used here are:

- Pressure characteristics
 - Socket (Man/Machine Interface)
 - Foot
- Elements of Shock Absorption
 - Impact Forces
- Work Requirements
 - Step-to-Step Transition Cost

Evaluating these parameters gives us a thorough assessment of amputee gait characteristics and how they are affected by the choice of prosthesis. Also, by using these parameters to compare amputees to non-amputees, we will gain insight into how close current prosthetic devices come to achieving “normal” walking behavior.

Analysis of amputee gait characteristics calls for us to conduct a set of clinical trials. First, we must recruit subjects suitable for our study with the appropriate amputation and activity levels. Then, we will conduct level ground walking experiments with amputees using their current passive prosthesis and an active bionic ankle, controlling for walking velocity and recording a variety of sensor data. We will analyze the collected data to determine how applied stress and pressure are affected by a more biomimetic functioning prosthesis. Finally, we will look for any consistencies in the data that can be correlated to pain and discomfort. Hopefully, we can gain some insight into how to design and tune prostheses to optimize comfort, effectively reducing the risk of short- and long-term risks associating with lower extremity amputations.

1.2 Significance of Study

The major contribution of this study is to the science of rehabilitation. It aims to underscore the importance of mechanical reproduction of biological joint behavior in the development of rehabilitative devices. By designing assistive devices that closely mimic the natural behavior of biology, natural energy consumption and biomechanics can be restored to disabled individuals. In the case of the PowerFoot, inertial measurements enable control of electromechanical components, which match the power output of the biological ankle joint. The same approach may be used in the design of rehabilitative devices to restore normal biological function to other joints.

Overall, this study offers insight into how to better design prosthetic and rehabilitative devices to minimize discomfort, which will eventually lead to prolonged use of the device and reduced risk of medical complications.

1.3 Summary of Chapters

Chapter 2 outlines several aspects of gait analysis that will aid in reading this thesis. It discusses leg anatomy, muscle function, and ambulation following a lower extremity amputation. The chapter also features a description of the current state of ankle prostheses.

Chapter 3 details the methodology undertaken to complete this thesis. This chapter offers a thorough description of the data collection and analysis protocol. It discusses all equipment and computational tools used for this study.

Chapter 4 describes the key findings of this project. Here, we lay out the results of this

study and discuss what conclusions can be extracted from the data we have collected.

In Chapter 5, we conclude this thesis, summarizing its findings and offering suggestions for future work.

Chapter 2

Background

This chapter outlines previous studies relevant for understanding and undertaking this project. It begins with a basic introduction to human gait analysis, defining key terminology in the field and describing major studies related to gait biomechanics, bipedal locomotion, and muscle physiology. Next, the chapter goes into pressure studies, with a specific focus on LEA and human walking. We will discuss important background research on pressure distribution during the gait cycle and how it affects common medical risks. Finally, the chapter ends with an overview of the current state of ankle prostheses.

2.1 Gait Analysis

This section discusses key aspects of human gait analysis. We will introduce commonly used terminology and concepts.

2.1.1 The Gait Cycle

The periodic behavior of legged locomotion is referred to as the gait cycle. Bounded by consecutive heel-strikes of a single limb, specific actions involved in walking are often discussed as a percentage of the total gait cycle from 0% to 100%. The human gait cycle can be subdivided into two segments, stance phase and swing phase. Each period/cycle is

bounded by consecutive heel-strikes of a single limb. Stance phase, in which the foot is in contact with the ground, accounts for 60% of the gait cycle. Conversely, swing phase accounts for the remaining 40% and represents the points in which the foot is not in contact with the ground. Both phases can be further subdivided to more clearly describe the behavior of the lower extremities during the gait cycle. Figure 2-1 is a clear depiction of the phases.

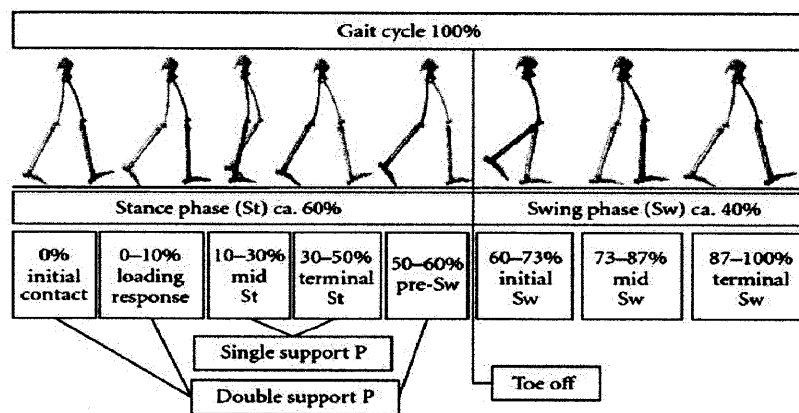


Figure 2-1: Phases of the Gait Cycle [Perry 2003]

Each limb experiences its own gait cycle. The right leg's cycle begins and ends with consecutive heel-strikes of the right leg. The opposite is true for the left leg. However, by observing the two cycles in relation to one another, two key concepts may be defined: single support and double support. Both terms have obvious definitions, as their names come from the number of legs supporting the body at a particular instance. Most often, the right and left legs are in opposite phases. One is in stance while the other is in swing (single support). However, between phase transitions, both feet touch the ground (double support). Familiarity with these two terms is required to understand a concept that will be discussed later in this thesis, step-to-step transition work.

The gait cycle is a crucial part of this, and any other, study involving walking. As one of the key concepts of gait analysis, walking is usually examined with respect to the gait cycle.

With this said, many of the plots to follow in this thesis will be projected as functions of gait cycle percentage.

2.1.2 Shock Absorption

Transference of body weight from the trailing leg to the leading leg occurs abruptly during double support (see Figure 2-1). Just before impact, the body is essentially free falling toward the ground until the foot of the leading leg contacts the floor, resulting in a load transfer of approximately 60% of the body weight to the leading leg in about 0.02 seconds [Perry 1992]. This process is known as load response and accounts for the first 10% of the gait cycle. Abrupt weight shifts are handled by the lower extremities through shock absorbing reactions at the ankle, knee, and hip. These reactions are illustrated in Figure 2-2.

At the end of terminal swing the body enters a short free fall state in which body weight shifts abruptly from the trailing leg to the leading leg (Figure 2-2a). At initial ground contact, the ankle immediately plantar flexes, slowing just before the forefoot touches the floor to reduce the rate at which body weight is distributed to the ground (Figure 2-2b). The knee joint acts as the greatest shock absorber during loading response, flexing at impact to absorb force. At this point, the quadriceps engage to accept some of the loading force; thus, reducing the load applied to the knee joint (Figure 2-2c). The last response to initial ground impact is a contralateral pelvic drop induced by a removal of support from the trailing leg's side of the pelvis (Figure 2-2d).

The behavior of each joint, as well as the corresponding muscles, plays a role in absorbing the immediate impact of body weight transfer between consecutive gait cycles. While the joints take on the majority of the load, the muscles act to reduce stresses applied to them.

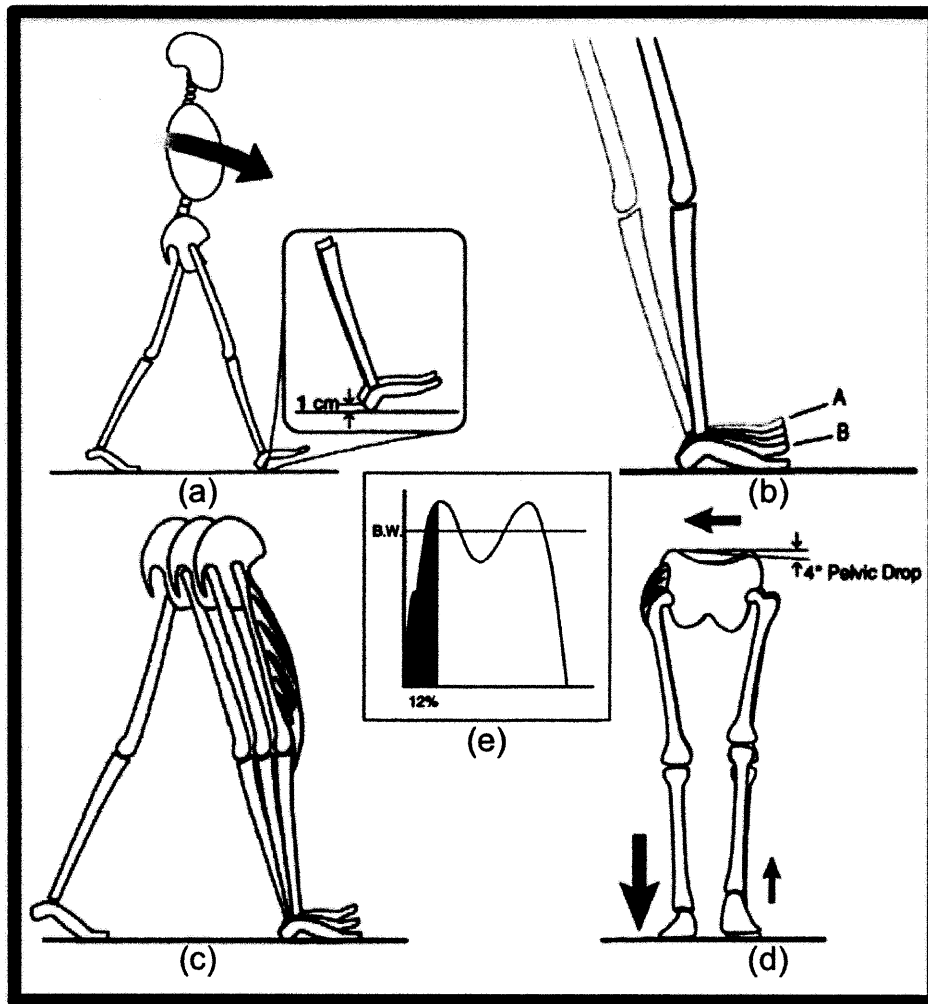


Figure 2-2: Shock absorption during loading response [Perry 1992]

2.2 Muscle Anatomy & Physiology

This section discusses human leg muscle anatomy and the role of leg muscles in walking. We will begin with an overview of leg anatomy and move into ways to characterize and analyze muscle activity during gait studies.

2.2.1 Anatomy of the Human Leg

Here, we will introduce the anatomy of the human leg, starting with a discussion of its skeletal structure, then giving an overview of its muscle-tendon units.

2.2.1.1 Skeletal Anatomy

There are three major bones that make up the human leg. These are the femur, tibia, and fibula. The femur and tibia connect to form the knee joint at the knee cap, or patella. We will use this junction, here and throughout this thesis, to reference regions of the lower extremity.

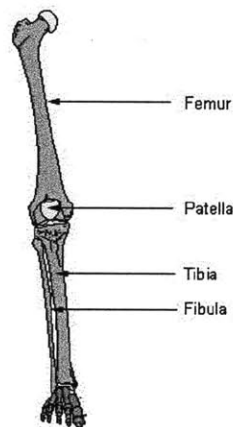


Figure 2-3: Diagram of Human Leg Bones [www.learnbones.com]

The femur makes up the region of the leg above the knee. Commonly referred to as the thighbone, the femur is the largest and longest bone in the body. In addition to forming the knee joint, the femur also connects with the pelvis at its acetabulum cup to form the hip joint [Whittle 2002].

Below the knee, the leg is comprised of two bones: the tibia and the fibula. The tibia is the larger of the two major bones below the knee. It is commonly referred to as the shinbone or shankbone, and forms the connection between the knee and the ankle. Running alongside the tibia is the fibula, which connects to the back of tibia head just below the knee joint and

extends below the tibia to form the lateral side of the ankle joint. The fibula is also called the calfbone, and is one of the most slender long bones in the body [Whittle 2002].

2.2.1.2 Muscular Anatomy

There are over 40 muscles in the human leg [Whittle 2002]. Each plays a very specific role in leg motion, assisting in the control of joint orientation and leg position. Leg muscles typically act in antagonistic pairs, with one muscle (agonist) generating movement in some direction and a second muscle (antagonist) generating movement in the opposite direction. Using this concept, leg muscles can be split into four different classes: extensors, flexors, adductors, and abductors. Flexors and extensors oppose one another and are thus an antagonistic set. Similarly, adductors and abductors oppose one another, forming a second antagonistic set.

The four classes of leg muscles are defined by the movement that they initiate about a joint. Flexors are muscles that generate a joint angle decrease in the sagittal plane [Whittle 2002]. They are opposed by extensors, which generate a joint angle increase in the sagittal plane, generally resulting in the straightening of the limb. Abductors are muscles that draw the limb away from the sagittal plane. This motion occurs in the coronal plane. Adductors oppose this motion, drawing the limb toward the sagittal plane. Each lower extremity joint has its own set of muscles that fall within at least two of these classes (some joints do not allow all four types of movement). Many times, a single muscle will contribute to mobility about more than one joint. The list of muscles in Table 2-1 shows this.

List of Leg Muscles			
Hip Abductors			
Gluteus Medius	Gluteus Minimus	Sartorius	
Hip Adductors			
Adductor Brevis	Adductor Longus	Adductor Magnus	Adductor Minimus
Pectineus	Gracilis	Obturator Externus	
Hip Extensors			
Gluteus Maximus	Bicep Femoris (Long head)	Semimembranosus	Semitendinosus
Hip Flexors			
Rectus Femoris	Sartorius	Adductor Longus	Adductor Brevis
Psoas Major	Psoas Minor	Iliacus	Gracilis
Pectineus	Tensor Fasciae Latae		
Knee Flexors			
Bicep Femoris	Sartorius	Gastrocnemius	Gracilis
Semimembranosus	Semitendinosus		
Knee Extensors			
Rectus Femoris	Vastus Intermedius	Vastus Lateralis	Vastus Medialis
Plantarflexors			
Gastrocnemius	Soleus	Flexor Hallucis Longus	Flexor Digitorum Longus
Plantaris	Tibialis Posterior	Peroneus Longus	Peroneus Brevis
Dorsiflexors			
Tibialis Anterior	Extensor Hallucis Longus	Extensor Digitorum Longus	Peroneus Tertius

Table 2-1: List of Leg Muscles [www.rad.washington.edu, Whittle 2002]

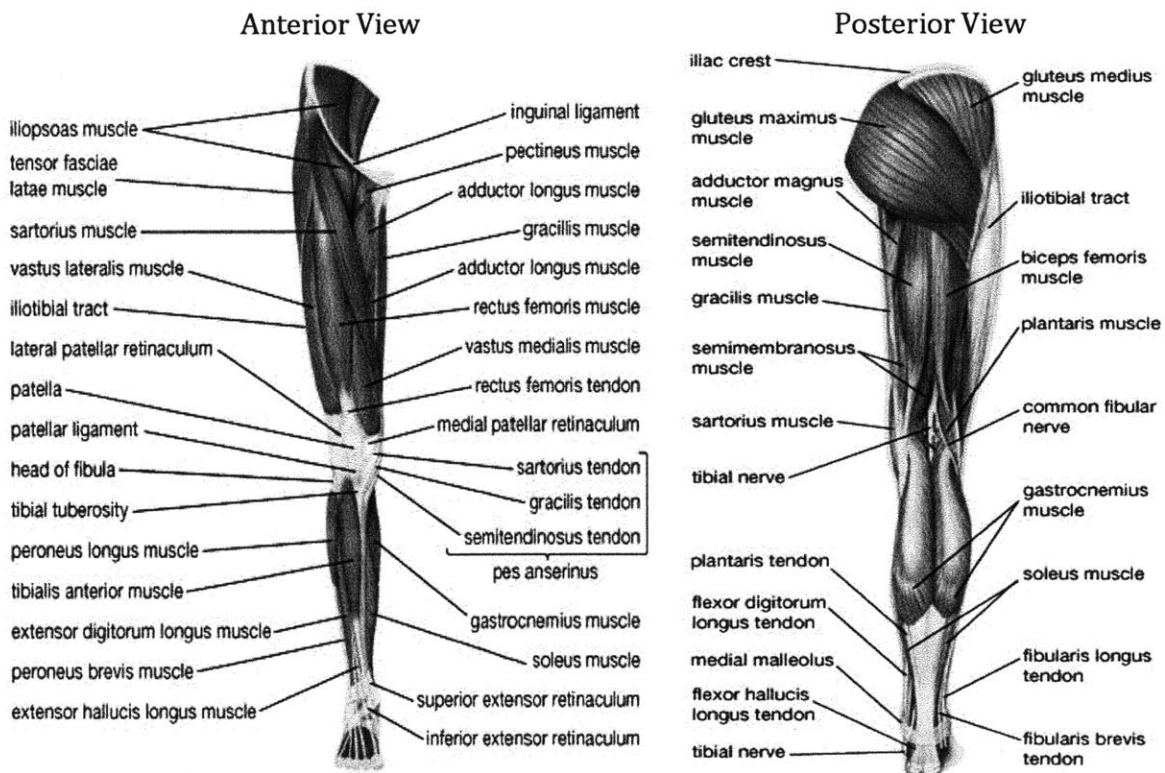


Figure 2-4: Leg Muscle Anatomy [www.britannica.com]

2.2.2 Muscle Activation

The best and most commonly used measure of neurological activation of skeletal muscle is electromyography. The following sections will define electromyography and discuss its common uses.

2.2.2.1 What is Electromyography?

Electromyography (EMG) is a commonly used approximation of neuromuscular activation. Peter Konrad, author of *The ABC of EMG: A Practical Introduction to Kinesiological Electromyography*, offers the following definition of electromyography:

Electromyography is an experimental technique concerned with the development, recording and analysis of myoelectric signals. Myoelectric signals are formed by physiological variations in the state of muscle fiber membranes [Konrad 2005].

Basically, electromyography is a measure of the electrical potential present in muscle bellies when they are neurologically activated. Inactive muscles produce no electric potential. When muscles contract, muscle fibers begin to produce action potentials, which are detected by an electromyograph. As muscles reach maximum contraction, more action potentials are produced, resulting in an increase in signal amplitude.

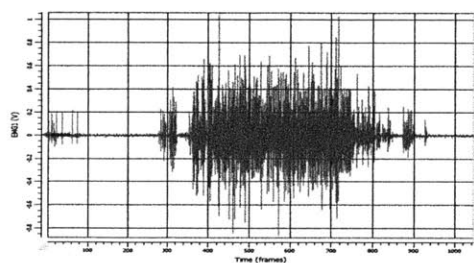


Figure 2-5: Electromyogram/EMG Signal

2.2.2.2 Common Criticisms

Electromyographic signals are a function of many different parameters. These parameters can either be related to the muscle or the electrode. In reference to muscle parameters, EMG signals are affected by fat surrounding the muscle, muscle temperature, and the cross-sectional area of the muscle belly. On the other hand, electrode parameters include the electrode's placement, size, and shape [Winter 1991]. Changes in each of these parameters create deficiencies in readings, which complicates EMG analysis.

Another criticism of electromyography is its lack of consistencies. EMG signals are highly unreliable because their voltage magnitudes vary between different subjects, in a single subject, and even in a single muscle. This extreme variance in signals makes processing nontrivial. Developing robust algorithms and making general inferences is nearly impossible. In some cases, the timing of EMG bursts is somewhat consistent and can be used as a good on/off indicator. This is particularly useful for deriving neuromuscular gait models [Geyer et. al 2010]. However, most applications require consistency in magnitude.

2.2.2.3 Types of Electrodes

There are two commonly used types of EMG electrodes: wire and surface. Both offer their own distinct advantages and challenges, and much work has gone into comparing the two over the years. Wire electrodes are invasive and can be painful, but they offer the most reliable readings from small and deep muscles [Winter 1991]. However, they are susceptible to cross talk from deep muscles. On the other hand, surface electrodes are the most commonly used and most reliable electrodes. There are drawbacks to surface electrodes. They are particularly vulnerable to motion artifacts and poor placement. These drawbacks can lead to incredibly noisy recordings and tarnished data. Recently, groups have begun looking into implanted electrodes, which could offer a solution to the flaws of other electrodes.

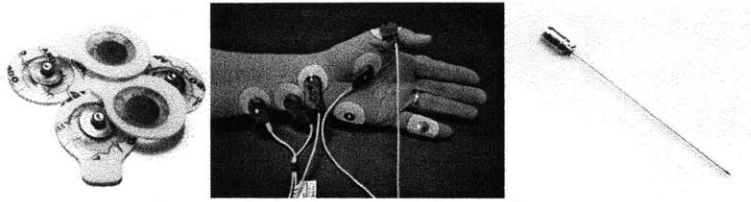


Figure 2-6: Types of Electrodes

The leftmost and middle images are of surface electrodes. The middle image shows them attached to a hand with the proper electronic connections to feed signals to the appropriate EMG signal reader. The rightmost picture shows a wire (invasive) electrode. [12, 13, 14]

2.2.2.4 Common Uses of Electromyography

EMG is used commonly in several different fields, including medical research, sports science, and rehabilitation. It yields information on muscle behavior and offers numerical feedback that is useful for training muscles [Konrad 2005]. Primarily, it is used as an evaluation tool in physiological and biomechanical studies. When EMG was originally introduced in the 17th century, it was used to analyze muscles in electric eels. The first use of EMG to monitor voluntary muscle contractions did not come until the late 19th century when Étienne-Jules Marey conducted experiments to analyze muscle contractions and subsequently coined the term electromyography [Cram et. al 1983]. Now, EMG has popped up in many new applications, in both science and engineering.

Gait analysts have adopted EMG for use in gauging muscle activation patterns during the gait cycle and developing human locomotion models. It helps determine biologically accurate neural activation, which is commonly used to derive realistic neuromuscular models [Geyer et. al 2010]. Additionally, recent advances in prosthetic devices have focused on incorporating neural control to help facilitate mobility. Control schemes for prostheses are being developed that monitor EMG and use it to implement communication between the wearer and the device [Wang 2010]. These schemes have been pursued more frequently in upper extremity devices, but instances of neural control with lower extremity devices are becoming more prevalent.

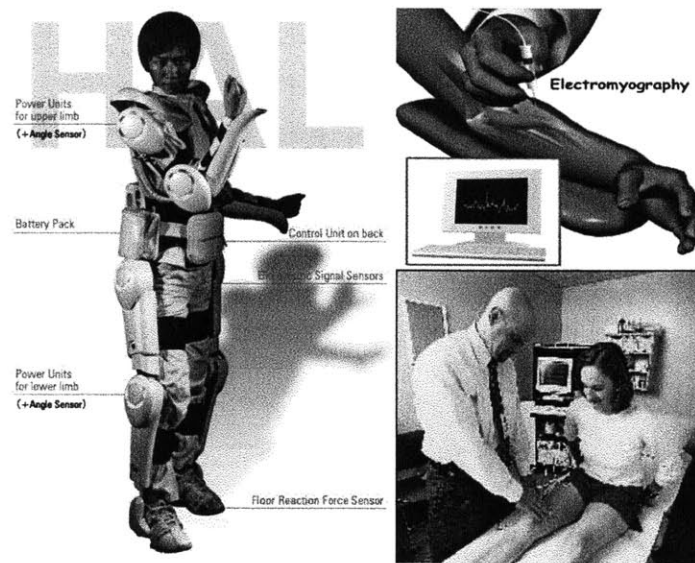


Figure 2-7: Uses of EMG

HAL is a robotic exoskeleton that uses electromyography to infer user intention and assist in power generation (left). The image in the top right corner is an animated depiction of a doctor using wire electrodes to monitor EMG in a patient's arm. The bottom right image shows a doctor using EMG to evaluate a patient's leg muscles. [16, 17, 18, 19]

2.2.2.5 EMG During the Gait Cycle

Analyzing electromyography profiles during the gait cycle offers important insight into the operations of leg muscles in locomotion generation. Observing these profiles gives a clear view of which muscles are active during certain points of gait. EMG profiles, which may be displayed in either microvolts or as normalized percentages, for eight key lower extremity muscles are shown in Figure 2-8. Do notice the standard deviations of the profiles. They show the inconsistencies that are often complained about in EMG studies.

Of the eight muscles displayed in Figure 2-8, six (Gluteus Maximus, Rectus Femoris, Vastus Lateralis, Vastus Medialis, Tibialis Anterior, and Lateral Hamstring) seem to act primarily during late swing, loading response, and early stance. The tibialis anterior, though, does show significant activity during early swing. This is the muscle that generates swing phase

dorsiflexion for ground clearance. The medial gastrocnemius acts only during late stance phase. It works to generate power plantarflexion right before swing. The remaining muscle, the adductor magnus works throughout the gait cycle to keep the leg toward the center line of progression.

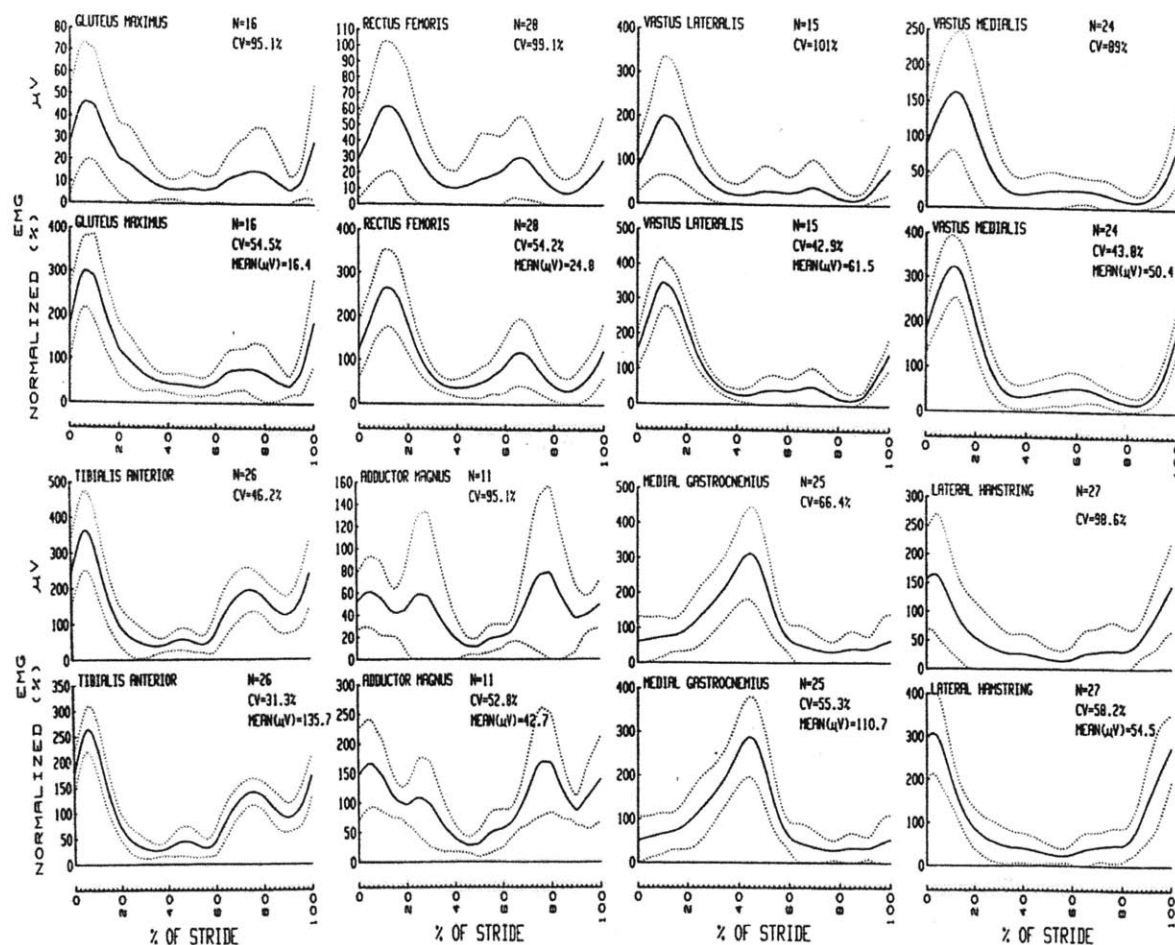


Figure 2-8: EMG Profiles during the gait cycle [Winter 1991]

2.3 Post-Amputation Gait

To this point, we have only dealt with gait characteristics pertaining to healthy able-bodied adults. However, this thesis primarily concerns itself with gait pathologies and behaviors

associated with persons living with a lower extremity amputation. Specifically, we will focus on below-knee amputees missing a portion of the leg only on one side of the body. Abnormalities in the lower extremities cause significant differences in gait. Lower extremity amputations are no exception.

Many in the past have studied deviations from normal gait, as a result of an amputation. Previous work has focused on changes in metabolic requirements of walking, joint dynamics, and kinematics. These characteristics also differ between prostheses; thus, further complicating amputee gait analysis. The following sections will highlight previous work in amputee gait analysis, showing how it differs from “normal” gait.

2.3.1 Amputation Levels

Lower extremity amputations occur at six different levels identified in reference to leg joints. These levels include:

- a) Hip Disarticulation/Hemipelvectomy
- b) Above Knee (Transfemoral)
- c) Knee Disarticulation
- d) Below Knee (Transtibial)
- e) Ankle Disarticulation (Symes)
- f) Partial Foot

Each amputation level presents its own complications, ranging from a shortage of effective prostheses to insufficient residual limb/prosthesis interface mechanisms. Obviously, amputations of the lower limbs cause difficulties in locomotion. Each of the levels listed above create deficiencies in energy expenditure and discomfort, albeit to a different degree. We will focus on transtibial amputee locomotion.

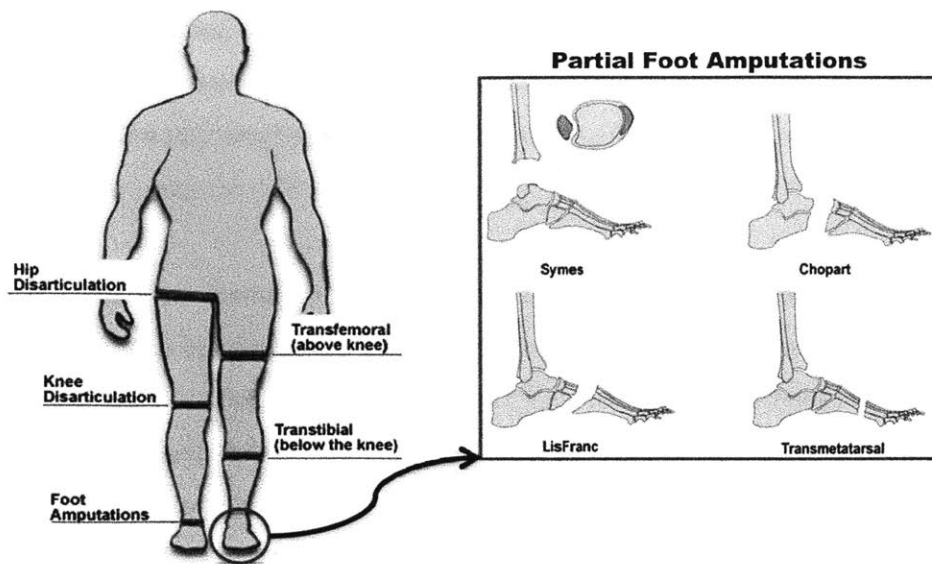


Figure 2-9: Amputation Levels of the Lower Limb [21, 22]

2.3.2 Transtibial Amputee Locomotion

Below Knee Amputees (BKA) exhibit substantial gait abnormalities resulting from the lack of a fully functioning ankle joint. The ankle plays key roles in walking - adjusting angle to accommodate any type of terrain and generating propulsion forces to keep the walker moving forward - that are lost or at least affected when someone uses a prosthetic ankle. Common deficiencies in amputee gait include higher metabolic demands, asymmetry in gait patterns and impact forces, and reduced walking speed [Perry 1992]. Deficiencies make amputee ambulation much more difficult and much less efficient.

The biological ankle is capable of performing multi-directional movement: extension, flexion, adduction, abduction, and rotation. State of the art commercial ankle prostheses have much more constricted movement. Passive prosthetic ankles offer some compliance that serves to simulate very limited flexion and extension. Quasi-passive and active prostheses perform less limited flexion and extension. However, current commercial prostheses offer absolutely no rotation.

Limited mobility of prosthetic ankle joints causes significant but immeasurable challenges for below knee amputees. Prosthesis wearers lose the ability to easily terrain adapt and change directions. Both of these tasks are accomplished simply with the biological ankle because it can move in multiple degrees of freedom while maintaining stability and producing power. Prostheses do not have this ability. In fact, ankle prostheses are optimal for sagittal plane motion and are quite rigid. This rigidity makes sudden terrain transitions and walking on varying terrains (i.e. rocks, snow, mud) extremely uncomfortable. Additionally, since prosthetic ankles operate most efficiently in sagittal plane motion, amputees find it easiest to walk in straight lines along smooth level surfaces. The lack of rotation creates problems for amputees when attempting to change directions.

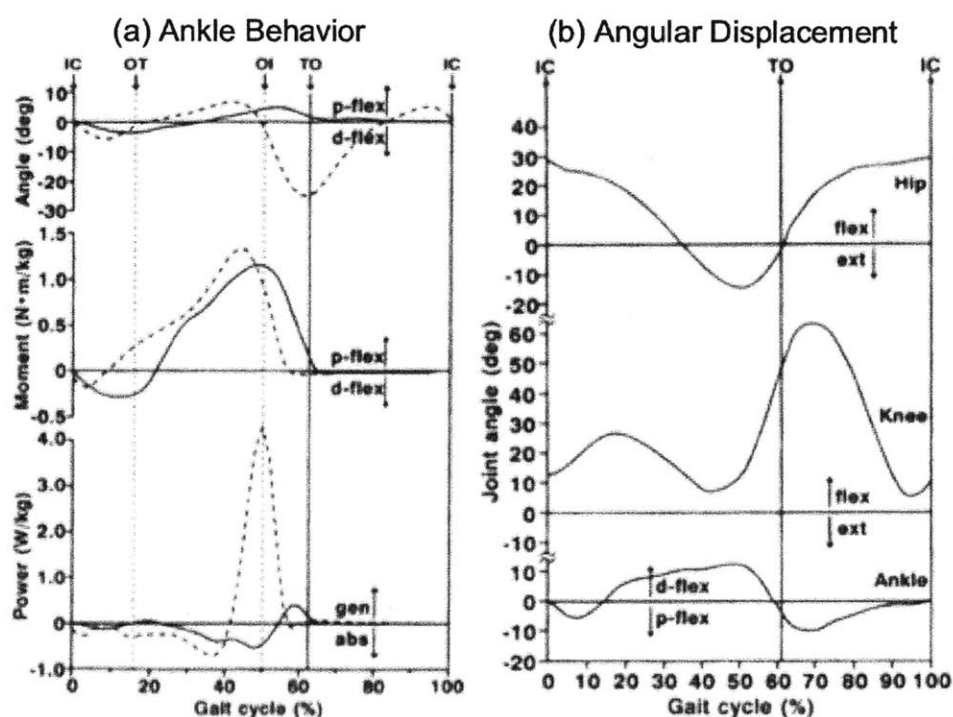


Figure 2-10: Joint Behavior of Transtibial Amputees [Whittle 2002]

Measurable results of below knee amputations include gait asymmetry, slower walking velocity, and higher metabolic expenditure. The absence of a power-producing ankle with

active plantarflexion capabilities creates timing and power deficiencies in each joint in the lower extremities, not just the ankle. However, Figure 2-10b shows nearly normal angular displacements of the hip, knee, and ankle. This means that even though transtibial amputees lack an active ankle joint, they can achieve center of mass displacement that closely resembles that of non-amputees [Saunders et. al 1953] due to compensation of the hip and knee. The ankle joint experiences the greatest changes in timing and dynamics of the BKA gait cycle (see Figure 2-10a). Most notably, the average prosthetic ankle joint appears to output an eighth of biological ankle power. Ankle absence puts a lot of pressure on the remaining joints to compensate for the loss of power.

2.3.3 Functional Levels

Transtibial amputees are classified based on their activity, or functional, level. This functional level determines which prosthetic components are chosen by the prosthetist for daily use by the amputee. The rating is determined by the type, duration, location, and difficulty of activities performed throughout a typical day. Once resolved, prosthetic components are chosen. Functional levels do change over time, increasing or decreasing with the activity levels of patients. In this situation, new components are chosen to better suit the updated requirements of the individual.

The five functional levels along with a description is included in the table below:

Functional Level 0	The patient does not have the ability or potential to ambulate or transfer safely with or without assistance and a prosthesis does not enhance his/her quality of life or mobility.
Functional Level 1	The patient has the ability or potential to use a prosthesis for transfers or ambulation on level surfaces at fixed cadence. Typical of the limited and unlimited household ambulator.
Functional Level 2	The patient has the ability or potential for ambulation with the ability to traverse low level environmental barriers such as curbs, stairs, or uneven surfaces. Typical of the limited community ambulator.

Functional Level 3	The patient has the ability or potential for ambulation with variable cadence. Typical of the community ambulator who has the ability to traverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic utilization beyond simple locomotion.
Functional Level 4	The patient has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels. Typical of the prosthetic demands of the child, active adult, or athlete.

Table 2-2: Functional Levels of Transtibial Amputees [www.yankebionics.com]

Functional levels are typically referred to using the letter “K” in place of “Functional Level”. Therefore, each functional level above may be referenced by K0, K1, K2, K3, and K4, respectively. PowerFoot users are typically of level K3 or above, representing individuals with fairly active lifestyles and the ability to perform quite complex physical activities.

2.4 Pressure Characteristics

Pressure distribution at the foot and socket levels are of great importance in this thesis. Here, we will discuss key previous studies of pressure characteristics in gait analysis for both amputees and non-amputees.

2.4.1 Foot Pressure

Monitoring pressure beneath the feet has been used in gait analysis for decades. Mostly, it is used to evaluate diabetic and arthritis patients whom exhibit higher than normal foot pressure. Typical foot pressures never exceed 1500 kPa, with pressure only reaching this level in sporting events. However, in diabetic neuropathy, recorded pressures have reached up to 3000 kPa [Whittle 2002]. Pressure this high greatly reduces blood flow to certain points of the foot, resulting in skin ulcerations. Examples of this have been shown by a Brazilian group, Bacarin et. al, that used plantar pressure distribution as a way to characterize diabetic patients with a history of foot ulcers [Bacarin et. al 2009].

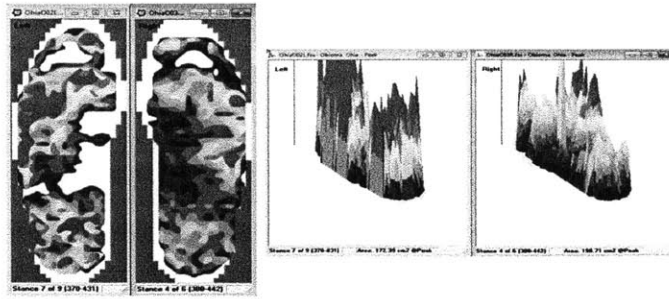


Figure 2-11: Foot Pressure Mappings

2.4.2 Socket Pressure

Prosthetic sockets remain one of the least developed parts of prostheses and one of the most common trouble spots. Even with technological advancements occurring in prosthetic component manufacturing, socket fitting and prosthesis alignment are still 100% reliant on man, the prosthetist. This creates very few commonalities in fit between amputees. Researchers have used pressure distribution to evaluate socket fit, paying close attention to the key areas where pain occurs most: bony protuberances, areas of muscle deterioration, and compliant skin spots [Dou et. al 2006, Wolf et. al 2009]. The inconsistencies in socket fit have been expressed through previous studies involving socket pressure. In 2009, Wolf et. al showed that absolutely no inter-subject consistencies existed in pressure data, which shows the difficulty of drawing general inferences from socket pressure [Wolf et. al 2009].

2.5 Step-to-Step Transition Cost

In 2002, Arthur Kuo's group at the University of Michigan introduced a concept known as step-to-step transition cost [Donelan et. al 2002]. Derived from a simple inverted pendulum walking model, transition cost is a measurement of the amount of work performed by both legs during transitions between steps. Now, it has become a regularly used measuring tool in gait analysis [Grabowski & Herr 2011].

Very little work is required to project a person's center of mass during the single support period of the level-ground gait cycle. In fact, the vast majority of work performed in walking occurs during double support. This is the step-to-step transition work. It is defined as the amount of work performed by an individual limb on the body's center of mass while transitioning between steps. In one step transition, both limbs generate work simultaneously. The leading leg applies negative work on the ground at impact, while the trailing leg applies positive work at toe-off to propel the center of mass forward. The magnitudes must be equal and opposite in order for someone to walk at a constant velocity.

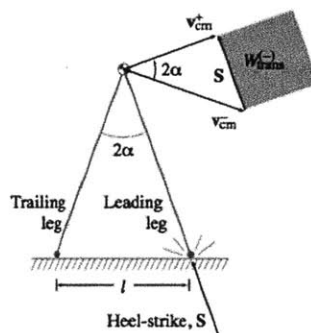


Figure 2-12: Step-to-step Transition Work [Kuo et. al 2005]

Step-to-step transition cost serves as a good indication of when one leg is being overexerted in comparison to the other. It can also be correlated to metabolics, with higher transition cost corresponding to increases in metabolic cost. Amputees walking with prostheses often require more transition work than normal. A recent study by Alena Grabowski and Huh Herr showed that transition cost was lowered by the use of a powered prosthesis [Grabowski & Herr 2011]. However, this study looked at transition cost as a whole and did not discuss the factors that may have contributed to the change. Art Kuo et. al proved that step transition work tends to increase with step length and width [Kuo et. al 2005]. Our study will provide a correlation between step length and transition work using a powered prosthesis.

2.6 Current State of Prosthesis Technology

The focus of recent innovation in prosthetic device designs has been to incorporate new forms of sensing, actuating, and control, to form a more seamless connection between man and machine and create intelligent interactions between the machine and its environment. Advancements in materials science have already led to the development of sophisticated passive prostheses, such as the commonly used Flex-Foot by Ossur Inc. [www.ossur.com], made of high strength, low weight, and high flexibility carbon fiber. Now, merging smart technology with these advanced materials has allowed for further development of prostheses, enabling them to sense their surroundings and act intelligently.

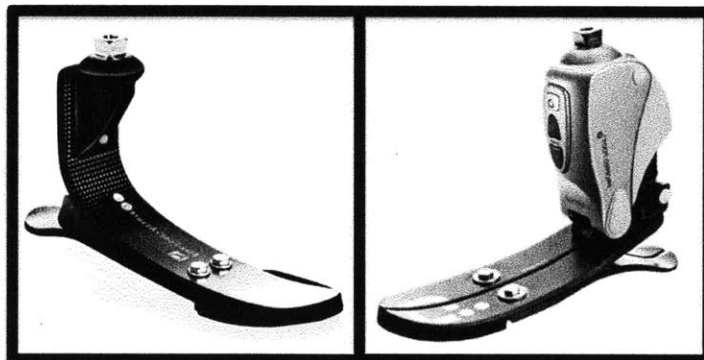


Figure 2-13: FlexFoot and ProprioFoot [www.ossur.com]

Biological limbs perform a number of tasks simultaneously to achieve the behavior needed to walk efficiently. Throughout the gait cycle, joints may act as dampers, power generators, or just flex and extend to ensure proper ground clearance. Lower limb prostheses must achieve these same functions while minimizing effects detrimental to gait efficiency. Therefore, “smart” lower limb prostheses must use proper electronics and sufficient mechanical components to perform simultaneous versatile tasks, but do so without exceeding natural limb weight and size.

No current lower limb prostheses match the highly tuned performance of the biological leg. However, a few prostheses are making strides. A quasi-passive ankle device, the Proprio Foot [www.ossur.com], uses sensors to determine the wearer's phase in the gait cycle. Then, using this information, the foot flexes or extends to facilitate ground clearance. Going a step further is the first active ankle device, the PowerFoot [www.iwalkpro.com], which uses inertial measurements and torque sensors to make informed decisions that control a series elastic actuator. The actuator, then, either generates positive work to propel the user forward or adjusts its angle to orient the foot correctly. Much like the biological ankle, these devices adapt, vary impedance, and vary power output depending on their stage in the gait cycle.

Present smart prostheses have formed what will become a new wave of bionic devices. As new sensing technology develops and as actuators become more sophisticated, prosthetic devices will come closer to achieving biological efficiency. Eventually, we will reach a day in which mechanical mechanisms and electronics will combine to produce devices that outperform biology.

2.6.1 PowerFoot Biom

The PowerFoot is the first powered ankle-foot prosthesis developed that employs both passive and active elements to more closely mimic the biological functions of the human ankle. Like the biological ankle, this device generates net positive work during stance phase and admits toe clearance during swing phase. The key feature of the PowerFoot is a series-elastic actuator comprised of a brushless motor and ballscrew transmission in series with a carbon composite leaf spring that stores and restores energy delivered by the motor, improving motor efficiency. Additionally, the powered ankle features a carbon-composite foot at the base of the prosthesis for added compliance, much like the Flex-Foot.



Figure 2-14: The PowerFoot Biom [www.iwalkpro.com]

The PowerFoot collects data from a variety of sensors to achieve biomimetic function, constantly varying ankle joint power output and stiffness throughout the gait cycle to match that of able-bodied individuals. Biologically inspired control systems govern the behavior of the device, enabling proper timing and magnitude of ankle power for a wide range of velocities. This sensing and adaptability is made possible by a six degree of freedom (three accelerometers and three gyroscopes) inertial measurement unit (IMU) and motor output encoders housed within the prosthesis. Encapsulated within a single housing is all electronics and a modular Lithium-Polymer battery to power the motor. The entire set-up sums up to a weight of approximately 2.0 kg. The weight of the prosthesis defines the user size range. It is known that the ankle-foot complex accounts for roughly two percent of a human's body mass. Therefore, the 2.0 kg powered ankle is appropriate to be worn by amputees with a pre-amputation body mass of around 100 kg.

Chapter 3

Experimental Methods

This thesis aims to uncover factors that influence pain and discomfort for transtibial amputees. To reveal these factors, we have designed walking trials, in which we will use subject feedback and sensor data to uncover extreme differences in gait biomechanics of transtibial amputees. This chapter outlines the details of the experiment. First, we will discuss the human trials conducted for this thesis, focusing on subject recruitment methods and evaluation tools we have used. Then, the chapter will end with a detailed description of the data analysis methods we have used to draw insight and form conclusions.

3.1 Subject Recruitment

Three unilateral below-knee amputees were recruited to participate in this study. Amputees will be recruited through a certified prosthetist that is familiar with their medical history and activity level. This insures that all participants are above a certain degree of disability and are able to complete all experiments. All amputee participants have no additional medical disorders and are at least K3 activity level.

3.1.1 Human Subject Use Approval

This study was pre-approved by the MIT Committee on the Use of Humans as Experimental Subjects (COUHES). Additionally, all subjects are required to sign consent agreements before participating.

3.2 Clinical Trials

This thesis is a complete study of amputee locomotion and how it is affected by prosthesis behavior. To evaluate this, we will conduct a series of trials with unilateral below-knee amputees walking on level terrain. Each amputee will complete the trials using their daily-use passive prosthesis and a PowerFoot Biom® active prosthesis at two different velocities (1.25 m/s & 1.75 m/s). We will compare the two devices based on biomechanical patterns, muscle behavior, and pressure characteristics.

Specifically, this study will focus on each device's effect on the elements of shock absorption, joint behavior, muscle activation, and pressure in the shoe and socket. All trials will be conducted in a motion capture facility complete with several sensors. Data analysis will be based on force plate, pressure sensor, electromyography, load cell, and marker trajectory measurements. The next section of this chapter discusses the equipment used for these measurements more in depth.

3.3 Equipment

The trials conducted in this study require subjects to be instrumented with a number of sensors. These sensors include:

- 1) VICON Motion Capture System
- 2) Delsys Wireless EMG System
- 3) Tekscan F-Scan Pressure System

The following sections will describe these instruments in more detail.

3.3.1 VICON Motion Capture System

The VICON Motion Capture System allows us to perform an in depth analysis of body motion. This system measures joint trajectories, and when combined with force plates, can be used to derive dynamics and kinematics. Twelve infrared cameras track strategically placed 12mm diameter IR reflective markers (see marker set description in the following section). Ground reaction forces are monitored using two AMTI force plates.

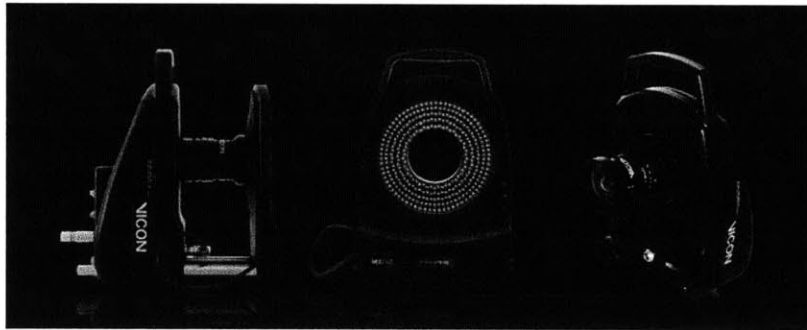


Figure 3-1: VICON Camera [www.vicon.com]

3.3.1.1 Marker Set

The marker set we use here is a modified Helen Hayes configuration. It is shown in Figure 3-2.

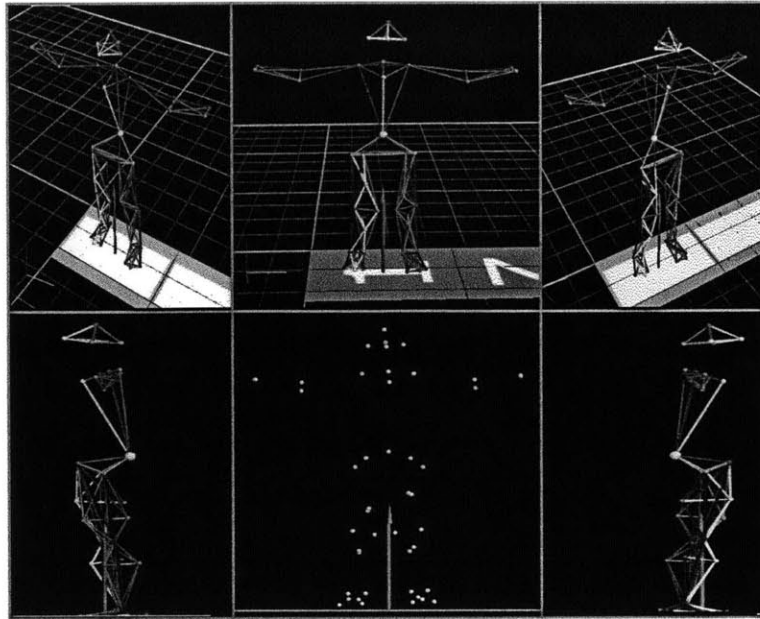


Figure 3-2: Marker Set

3.3.2 Delsys Wireless EMG System

To monitor muscle activation, we will use the Trigno Wireless System by Delsys. This system uses state-of-the-art wireless units that double as electromyography and motion sensors. The entire system consists of 16 EMG channels and 48 accelerometer channels (triaxial accelerometry in each sensor). Given the dual capabilities of these units, they run a little larger than typical surface EMG electrodes at 37mm x 26mm x 15mm. The Trigno system operates as a standalone device or as an input to compatible acquisition tools, and it can handle sampling rates of up to 4000 Hz.

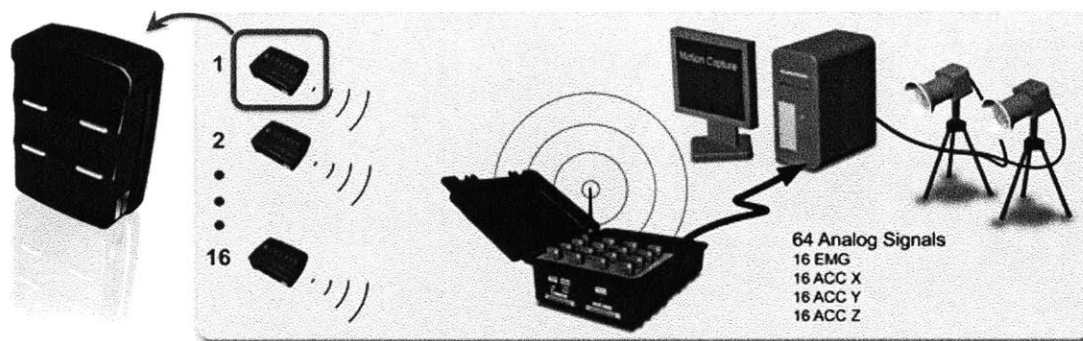


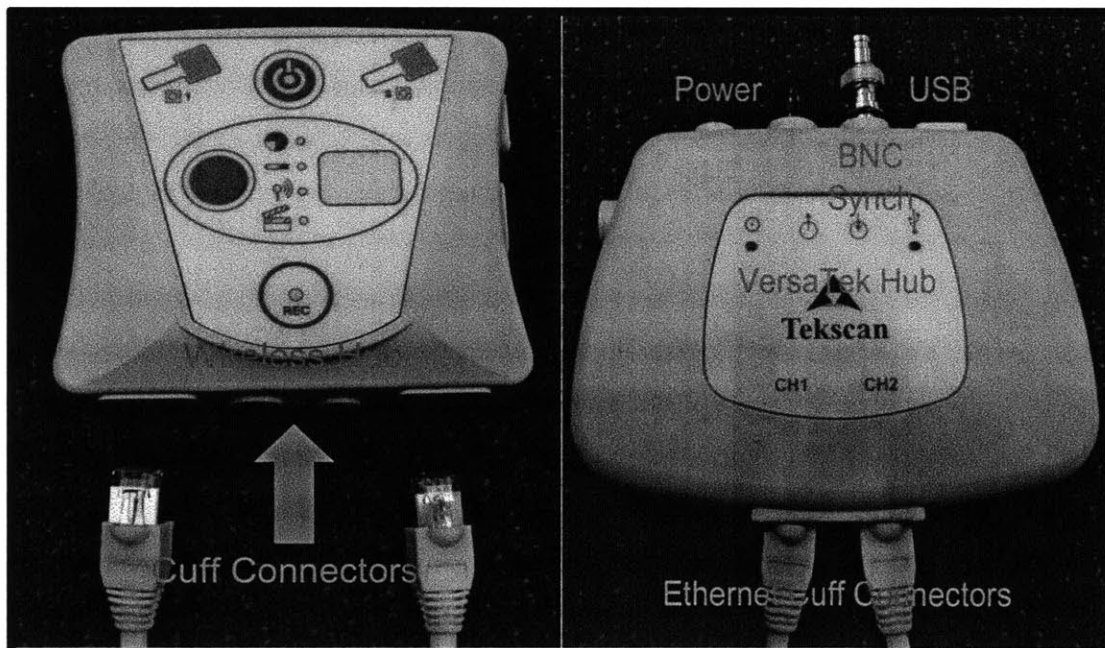
Figure 3-3: Trigno Wireless System [www.delsys.com]

3.3.3 Tekscan F-Scan Pressure System

Tekscan Incorporated, a South Boston company that specializes in pressure/force measurement systems, develops the F-Scan Pressure Mapping System for collecting real-time pressure characteristics. The F-Scan System is made specifically for gait studies. It comes equipped with thin flexible shoe insoles for mapping pressure distribution along the foot. The system can even be used for amputee gait trials since Tekscan produces an add-on that measures pressure within the human-prosthesis interface.

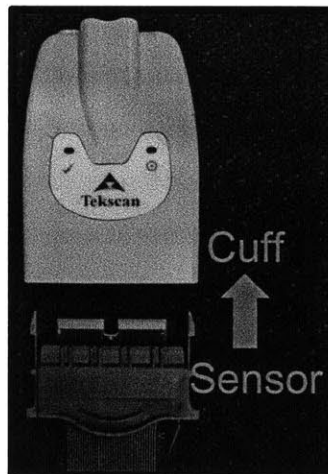
Tekscan's F-Scan Pressure Mapping System comes in both tethered and wireless setups. The systems are not only differentiated by the use of wires, but also in achievable sampling rates and synchronization methods. The wireless system is battery powered and transmits a WiFi signal from the sensor hub to the computer at a maximum sampling rate of 100Hz. The tethered system, on the other hand, is AC powered, connected to the computer via USB, and samples at a max rate of 750Hz. Using Tekscan's system in conjunction with other pieces of equipment becomes a challenge with the wireless system, as it does not offer wireless synchronization capabilities. The tethered unit includes BNC input/output channels for sending and receiving triggering signals. This makes it the more useful system when many tools are to be used simultaneously.

Both the tethered and wireless systems use what are called Versatek Hubs to facilitate communication between the sensors and the computer. The Versatek hubs are two-port units that receive information from at most two sensors through ethernet cables. The hub then transmits this information to the computer either through WiFi or USB. The system uses “cuffs” to allow communication between sensors and the hub. Ethernet cables run from one end of the cuff to the Versatek Hub. The opposite end of the cuff acts as the sensing sheet receiver, allowing the sensors to plug in directly and securing them in place.



a) Wireless Versatek Hub

b) Tethered Versatek Hub



c) Versatek Cuff

Figure 3-4: F-Scan Pressure Mapping System

This project uses two different sensors for monitoring pressure along the foot and the residual limb of below knee amputees. Both sensors are thin flexible sheets that can be altered to fit various sizes, shapes, and contours. The following sections describe the specifications for both sensors.

3.3.3.1 Foot Sensing

Tekscan developed the F-Scan System specifically to measure pressure characteristics of feet. The system comes equipped with easily insertable shoe sensors and is designed for optimal attachment to the ankle. Foot sensors come in the form of trim-able shoe insoles that can be cut to the size of any foot, size 13 (Men) or below. These sensors are thin and flexible enough to slide into the shoe without causing any drastic changes in fit and comfort. The specifications and schematic for the sensing insoles are shown in Figure 3-5.

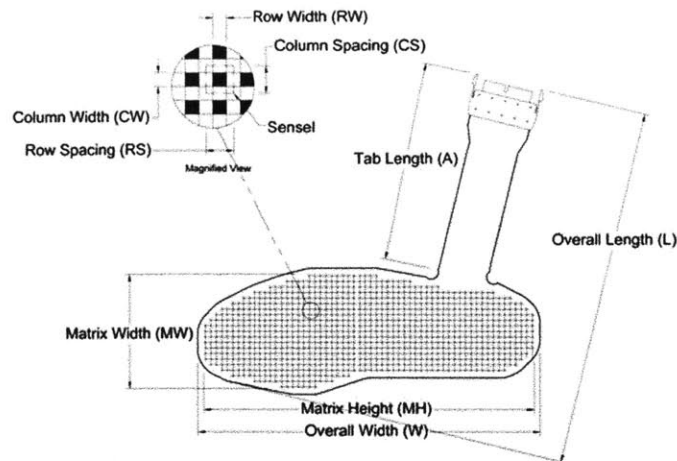


Figure 3-5: 3000E Sensor Diagram [www.tekscan.com]

Sensor attachment and security is critical to preserving the sensors and obtaining accurate recordings. To attach them, we use thin clear double-sided tape to adhere them to the bottom of the shoe. Before taping, the insoles are trimmed so that there is no bunching along the edges or curling up the sidewalls of the shoe. Enough tape is applied to minimize any shifting or creasing. In addition, subjects wear socks to reduce friction between their foot and the sensor. Any sensor movement or folds beneath the foot will contaminate recordings.

Overall Length L	Overall Width W	Tab Length A	Matrix Width MW	Matrix Height MH	Total Thickness	Pressure Ranges
12.88 in 327.1 mm	12.35 in 313.7 mm	7.19 in 182.6 mm	4.20 in 106.7 mm	12.00 in 304.8 mm	0.007 in 0.200 mm	75-125 psi 517-862 kPa

Columns			Rows			Total No. of Sensels	Resolution Sensel Density
CW	Pitch CS	Qty.	RW	Pitch RS	Qty.		
0.100 in 2.5 mm	0.200 in 5.1 mm	21 21	0.100 in 2.5 mm	0.200 in 5.1 mm	60 60	954 954	25.0 sensels per in ² 3.9 sensels per cm ²

Table 3-1: 3000E Sensor Specifications [www.tekscan.com]

3.3.3.2 Socket Sensing

An add-on can be purchased for the F-Scan System that allows it to be used for measuring pressure characteristics within the prosthetic socket. The add-on is called the F-Socket VersaTek Prosthetic Pressure Measurement System. Basically, it includes a different type of sensor and an additional sensor map for displaying recordings in the Tekscan software. These sensors, however, are more basic than the foot sensors. They are merely flexible square sheets that can also be trimmed to fit inside the socket. The specifications and schematic for the socket sensing sheets are shown in Figure 3-6.

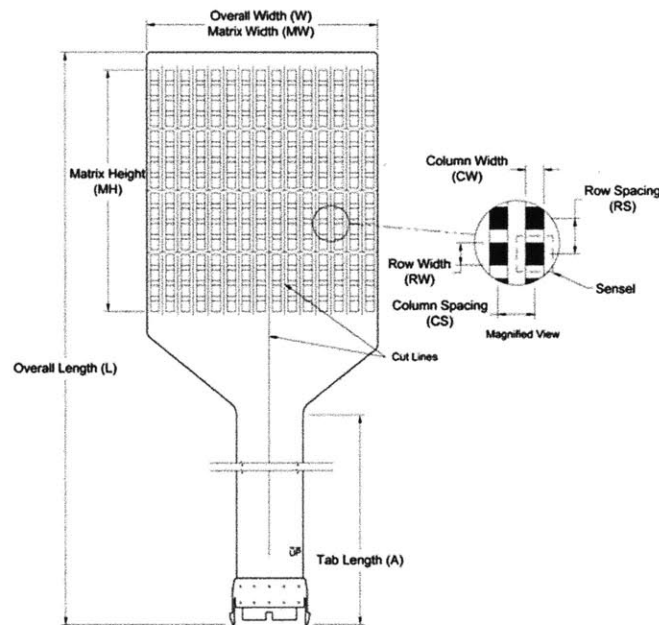


Figure 3-6: 9833E Sensor Diagram [www.tekscan.com]

Attachment of these sensors is also critical but nontrivial. They present a little more of a problem than the foot insoles. Similar to the other sensors, we use thin strips of double-sided tape to fix the sensors in place. However, the organic contour of sockets makes it a lot harder to fix the sensors in a way that eliminates movement and creases. We have found that the best option for us is to attach the sensors directly to the liner and then slide the liner into the socket.

Overall Length L	Overall Width W	Tab Length A	Matrix Width MW	Matrix Height MH	Total Thickness	Pressure Ranges
21.45 in 544.8 mm	7.50 in 190.5 mm	9.68 in 245.9 mm	7.50 in 190.5 mm	8.00 in 203.2 mm	0.007 in 0.200 mm	50 psi 345 kPa

Columns			Rows			Total No. of Sensels	Resolution Sensel Density
CW	Pitch CS	Qty.	RW	Pitch RS	Qty.		
0.265 in 6.7 mm	0.500 in 12.7 mm	15	0.350 in 8.9 mm	0.500 in 12.7 mm	16	240	4.0 sensels per in ²
		15			16	240	0.6 sensels per cm ²

Table 3-2: 9833E Sensor Specifications [www.tekscan.com]

Complete coverage of the socket walls requires two sensors. One sensor covers the front and sides of the socket and the other covers the back. Only the bottom cup of the socket, beneath the stump, is left exposed. With approximately 95% coverage area, we are able to analyze distinct regions of nearly the entire socket. By extracting data from specific areas along the sensor, we can pinpoint regions of high pressure corresponding to certain muscles or the unique form of the residual limb. Having this ability offers insight into the uniformity of pressure distribution within the socket and could clue us in to some common trouble spots.

3.4 Data Analysis

Analysis of data in this study will take place in a variety of channels. With so many sensing agents and other pieces of equipment, much of the data processing will take place in the software supporting individual pieces of hardware (Tekscan and VICON). In addition to these, we plan to use the SIMM Biomechanics Software Suite, which allows users to upload motion capture data to create musculoskeletal models and extract joint dynamics. Final processing and synchronization of all collected data will take place in a software package produced in MATLAB. The three key analyses that we will perform for this thesis are pressure distribution, shock absorption, and step-to-step transition cost.


Chapter 4

Results & Discussion

This chapter discusses the key findings of this thesis. We have used the techniques described in the Methodology section to build an experiment to uncover shock absorption and pressure changes spawned by the use of a more biomimetic ankle prosthesis. In many of the plots below, there will be identically colored thin lines outlining the main plot trajectories. These symbolize the standard deviations of the plots.

4.1 Subjects

We recruited three K3 level unilateral transtibial amputees to participate in this study. A description of each amputee participant is shown in Table 4-1.

	Subject ID	Age	Height	Weight	Amputation Side	Prosthesis Type	Suspension
Subject #1	JG	26	5'9"	165	Left	<div>Silhouette </div>	Suction



Subject #2	CC	49	5'11"	190	Left	Vari Flex 	Pin
Subject #3	AM	38	5'8"	150	Right	Renegade 	Suspension Sleeve

Table 4-1: Subject Information [31, 36, 37]

4.2 PowerFoot Tuning

The PowerFoot Biom is tuned via a Bluetooth connection between the prosthesis and an Android tablet or PDA. This connection allows us to control the power, timing, and stiffness of the PowerFoot. Each of these parameters differs for all users based on each individual's gait dynamics, tendencies, and comfort level. Tuning is also loosely based upon "normal" biomechanics, as defined by the norms found in healthy non-amputee walkers.

4.2.1 Tuning Parameters

Tweaking eleven parameters controls ankle power, timing, and stiffness. The values of each determine the collective behavior and feeling of the PowerFoot. Tuning parameter values for each subject are shown in Table 4-2. Pay close attention to the similarities and differences between each subject.

Subject ID	Set Weight	Pitch Velocity Threshold	Ankle Stiffness Percent	Ankle Stiffness Value	Ankle Stiffness Offset
JG	140	-250	55.5	855	5
CC	164	-300	69.5	995	5
AM	124	-250	71.2	1012	5

Power Percent	Timing Exponent Percent	Slow Walk Power Percent	K3 Bias	Toe Strike Percent	Toe Strike Enabled
80.0	0.00	50	20	66.6	100
40.3	100	3	0	66.6	100
71.3	60.3	49.6	10	66.7	0

Table 4-2: Subject Tuning Parameters

The PowerFoot Biom offers an unprecedented amount of adaptability. The eleven parameters shown in Table 4-2 allow us to create an optimized and custom feeling for each subject.

4.2.2 Discussion

The typical tuning process begins with stiffness tuning until the prosthesis feels stable and no longer flops excessively at ground impact. Next, we tune power output until it reaches a comfortable value for the wearer. It should not be too high or too low, and feel natural to the user. At this point, we can begin to tweak the timing until gait looks and feels optimal. Finally, we can play with the slow walk and toe strike parameters to account for certain tendencies of the user.

4.3 Step-to-Step Transition Work

Transition Work is a measure of how much work is required by each leg during the transitions between steps. We have calculated transition costs for each subject and will

compare them based on the type of prosthesis the subject is wearing. Figures 4-1 and 4-2 show this comparison.

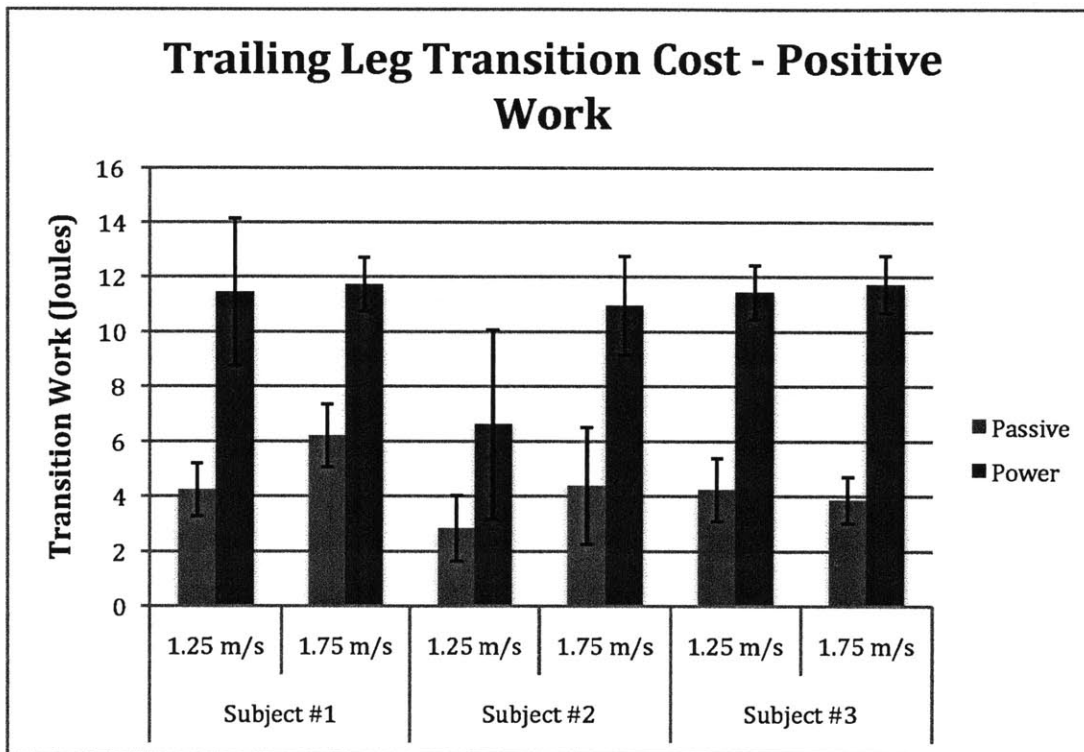


Figure 4-1: Trailing Leg Transition Cost

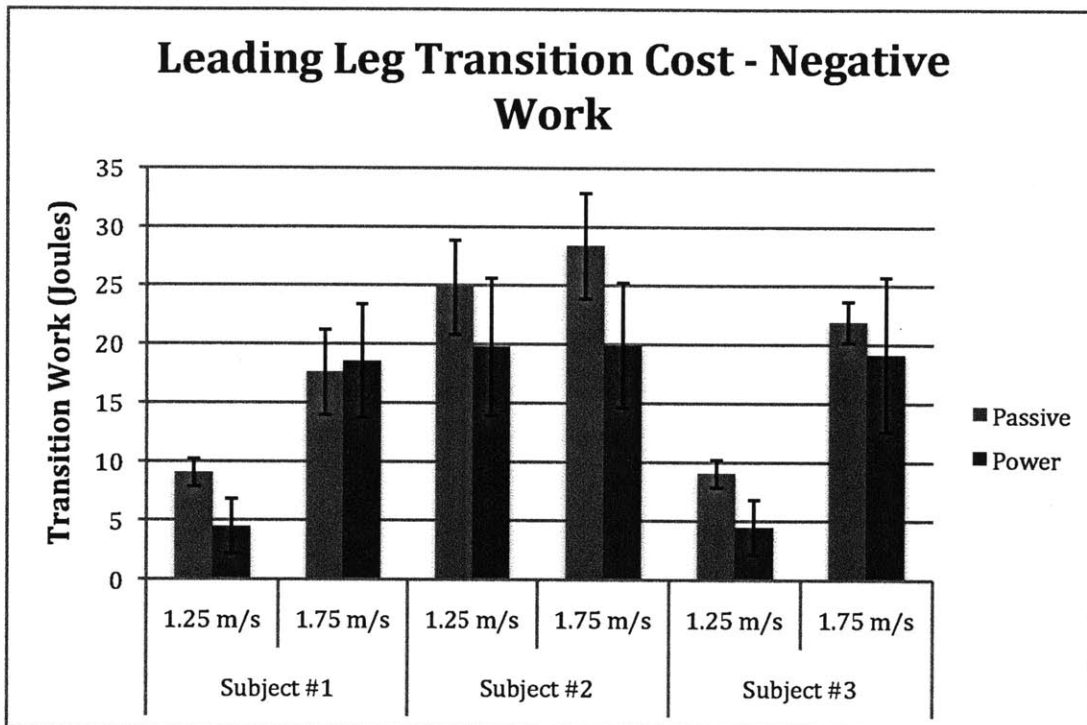


Figure 4-2: Leading Leg Transition Cost

4.3.1 Discussion

Transition work tells an interesting story about passive and active prostheses. We can see pretty clear trends in the charts above. In the trailing leg, there are consistent large increases in transition work of the power prosthesis equaling two or three times the amount of the passive prosthesis, in some cases. This increase is due to the increased power output of the prosthesis. To redirect the center of mass, the trailing leg must do positive work, and the powered ankle just has the ability to perform a greater amount of work. This increase in trailing leg work, theoretically, should reduce the amount of negative work the leading leg must do to make up for the deficiencies of the trailing leg. Let's see.

Figure 4-2 displays the negative step-to-step transition work performed by the leading leg. It shows, in general, a decrease in work needed when the amputee subjects walked using the powered ankle prosthesis, with 1.75 m/s for Subject #1 being the only exception. Subject #2 experienced the greatest changes, while Subjects #1 and #3 experienced smaller changes. The differences in the subjects are more than likely due to Subject #2 being within the recommended size range for the PowerFoot. Subjects #1 and #3 are below the recommended size and weight.

The increase in trailing leg work and decrease in leading leg work produce more normalized work requirements. Over time, this means that the unaffected limb will have to compensate to a far less degree.

4.4 Leading Leg Axial Force

We will now compare the ground reaction forces produced by each subject when switching between the power and passive prostheses. Our major focus in this section will be on the leading leg, which in this case stands for the unaffected limb or the limb opposite amputation. We want to see how the prosthesis affects its counterpart.

4.4.1 Subject #1

4.4.1.1 1.25 m/s

Trailing (Amputated) Leg

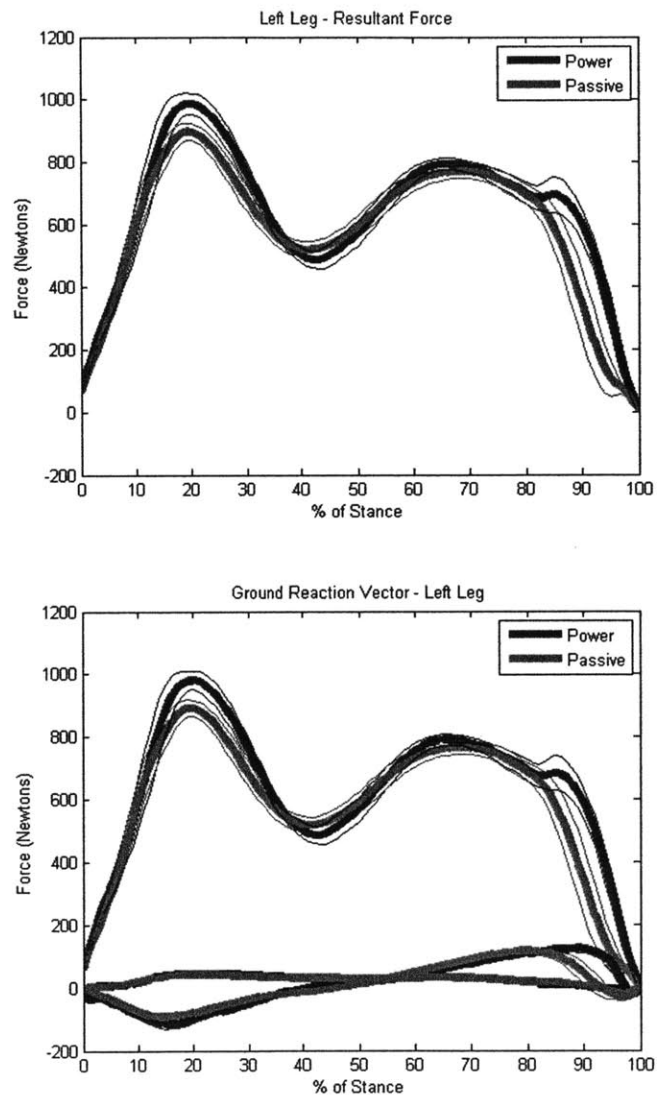


Figure 4-3: Ground Reaction & Axial Forces of Trailing Leg – Subject #1 (1.25 m/s)

Leading (Unaffected) Leg

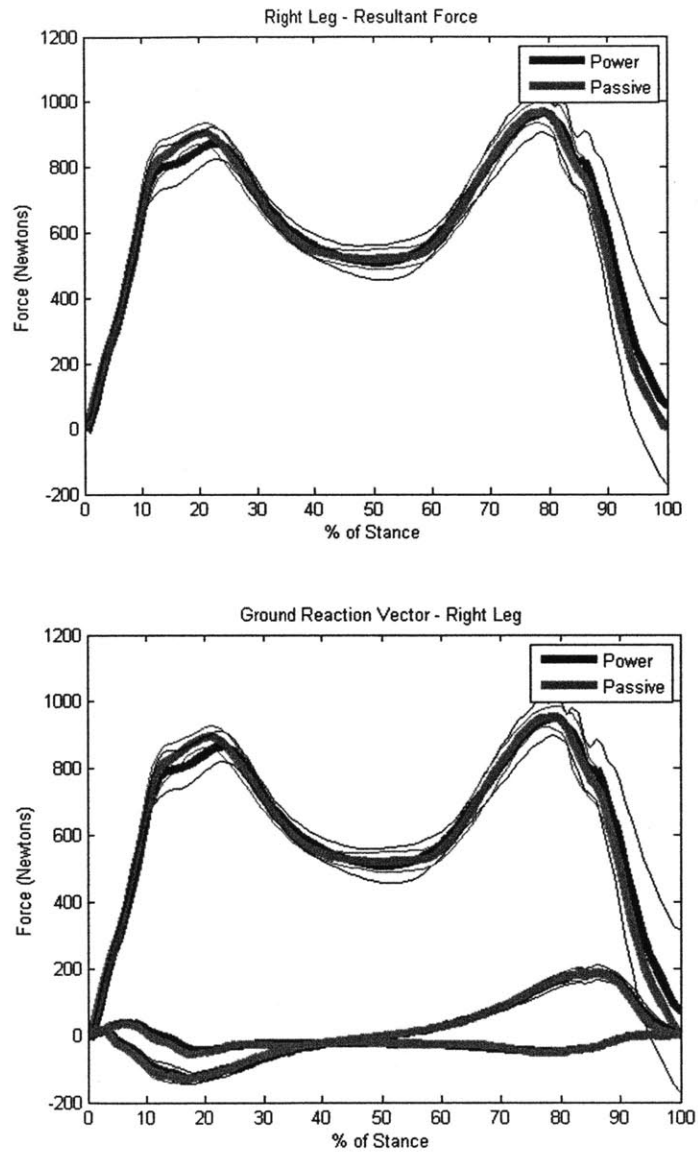


Figure 4-4: Ground Reaction & Axial Forces of Leading Leg – Subject #1 (1.25 m/s)

4.4.1.2 1.75 m/s

Trailing (Amputated) Leg

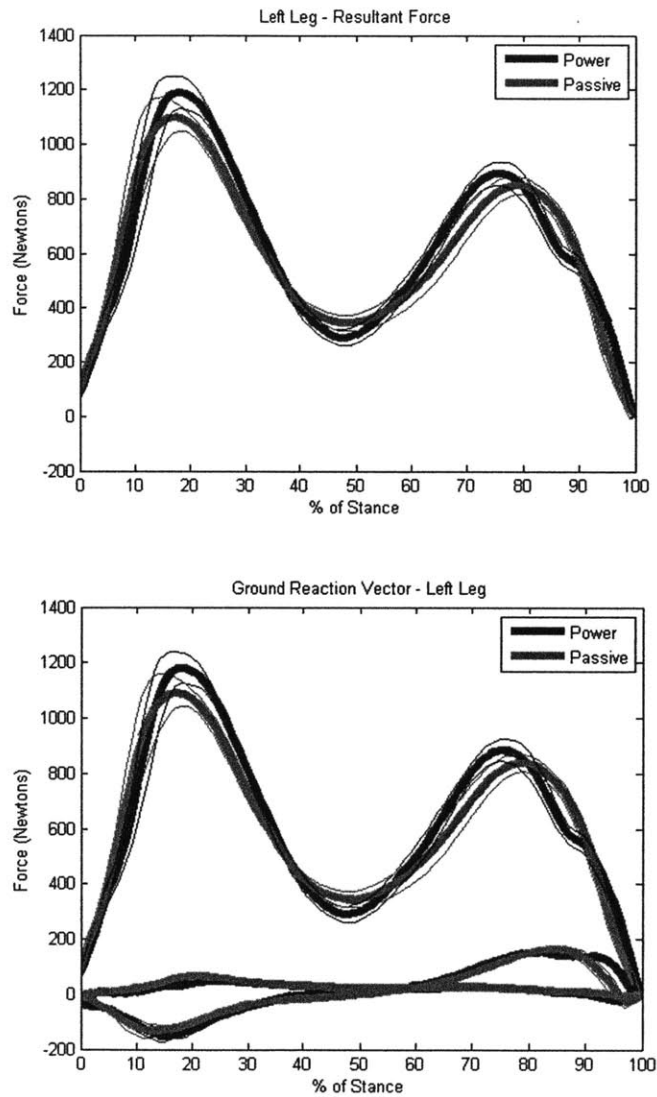


Figure 4-5: Ground Reaction & Axial Forces of Trailing Leg – Subject #1 (1.75 m/s)

Leading (Unaffected) Leg

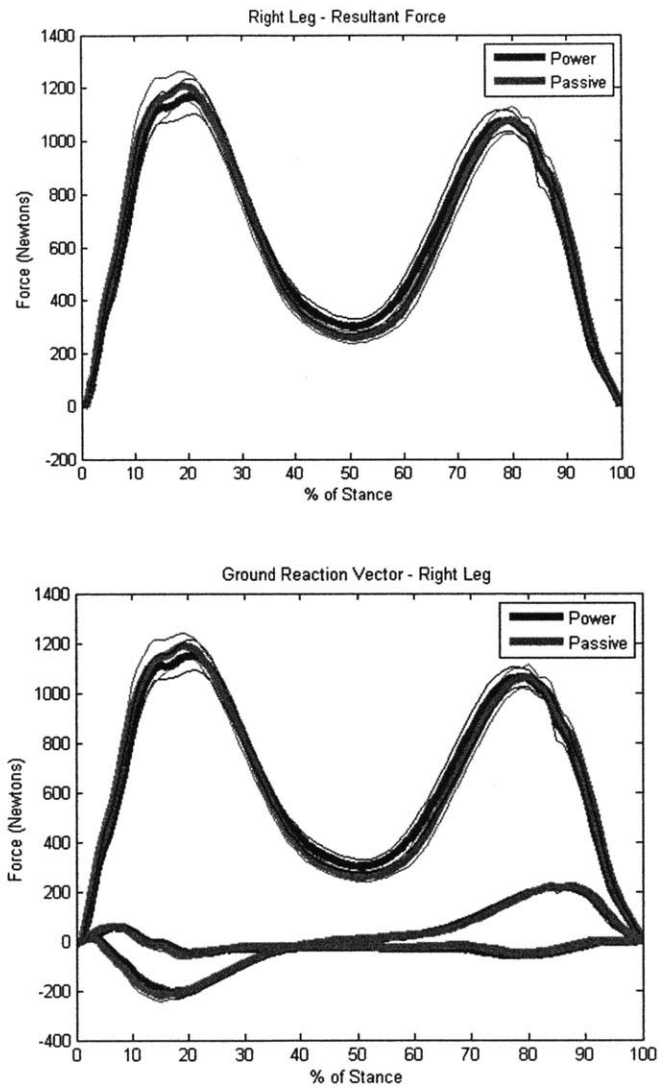


Figure 4-6: Ground Reaction & Axial Forces of Leading Leg – Subject #1 (1.75 m/s)

4.4.2 Subject #2

4.4.2.1 1.25 m/s

Trailing (Amputated) Leg

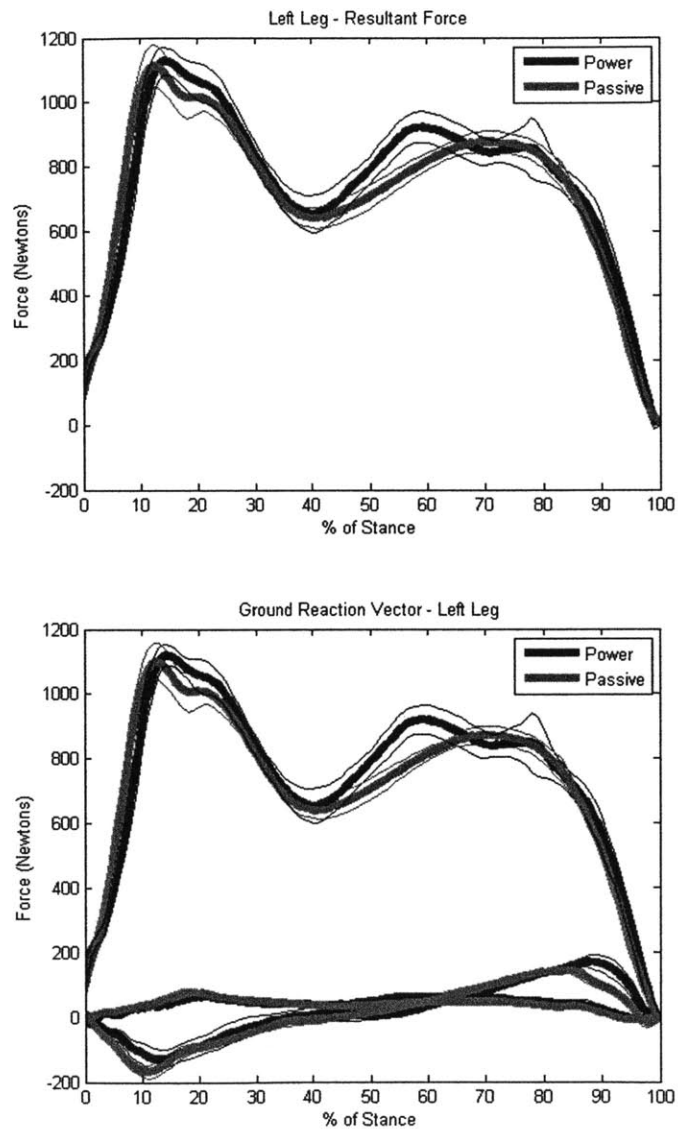


Figure 4-7: Ground Reaction & Axial Forces of Trailing Leg – Subject #2 (1.25 m/s)

Leading (Unaffected) Leg

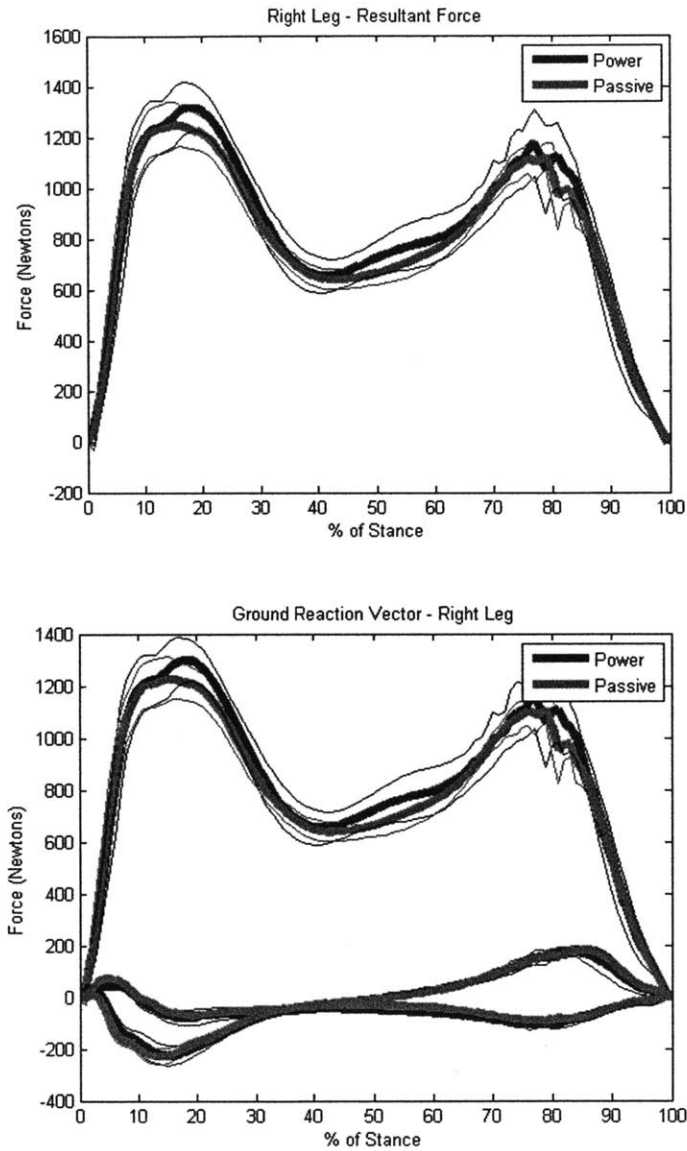


Figure 4-8: Ground Reaction & Axial Forces of Leading Leg – Subject #2 (1.25 m/s)

4.4.2.2 1.75 m/s

Trailing (Amputated) Leg

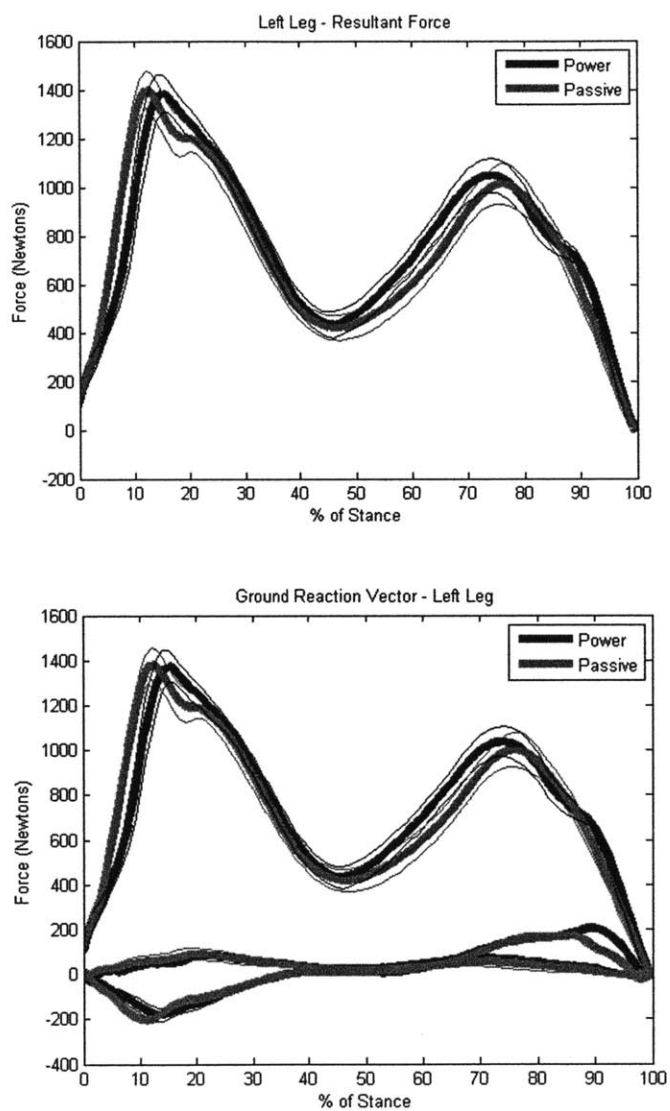


Figure 4-9: Ground Reaction & Axial Forces of Trailing Leg – Subject #2 (1.75 m/s)

Leading (Unaffected) Leg

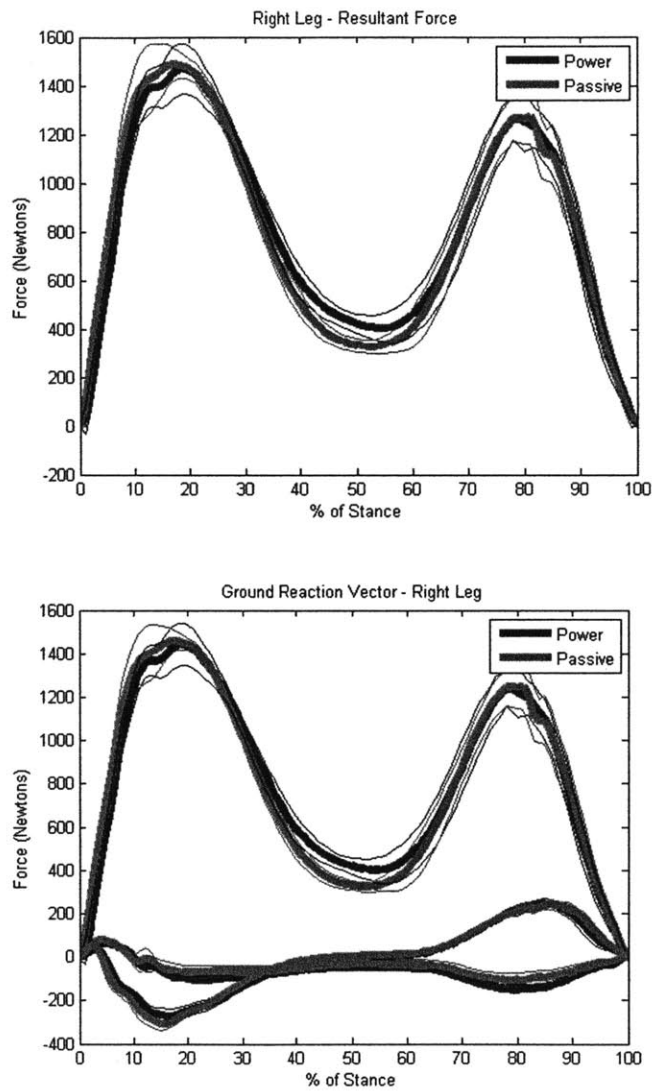


Figure 4-10: Ground Reaction & Axial Forces of Leading Leg – Subject #2 (1.75 m/s)

4.4.3 Subject #3

4.4.3.1 1.25 m/s

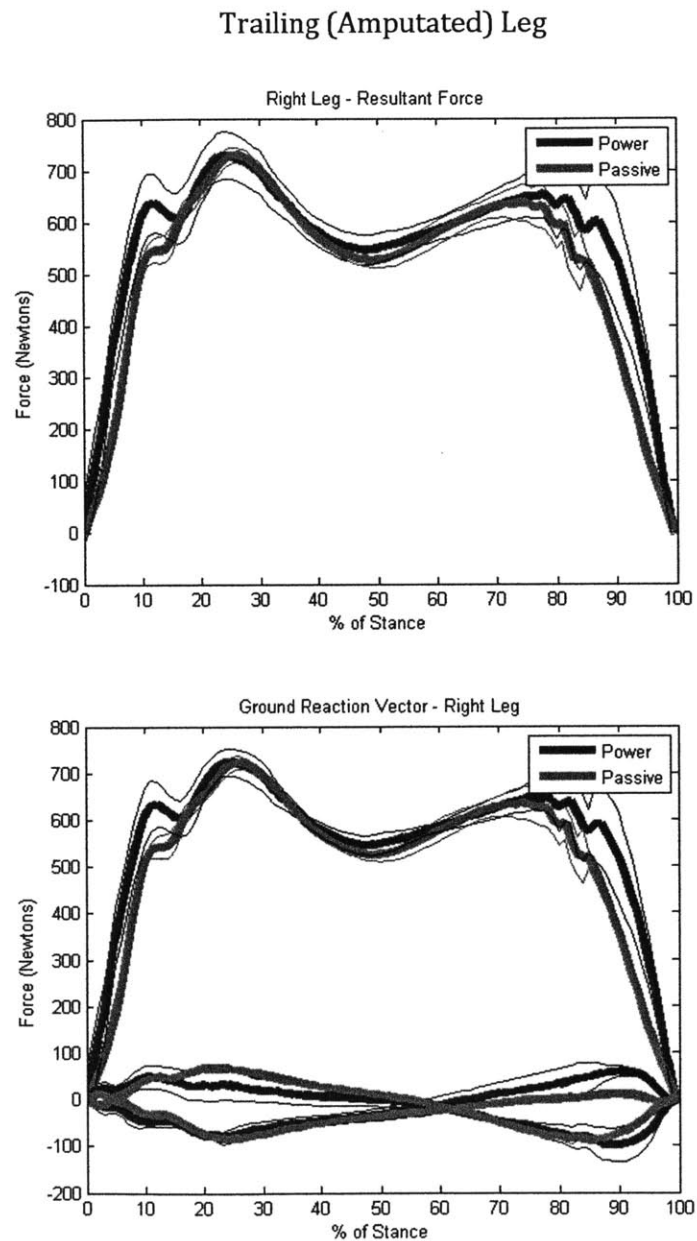


Figure 4-11: Ground Reaction & Axial Forces of Trailing Leg – Subject #3 (1.25 m/s)

Leading (Unaffected) Leg

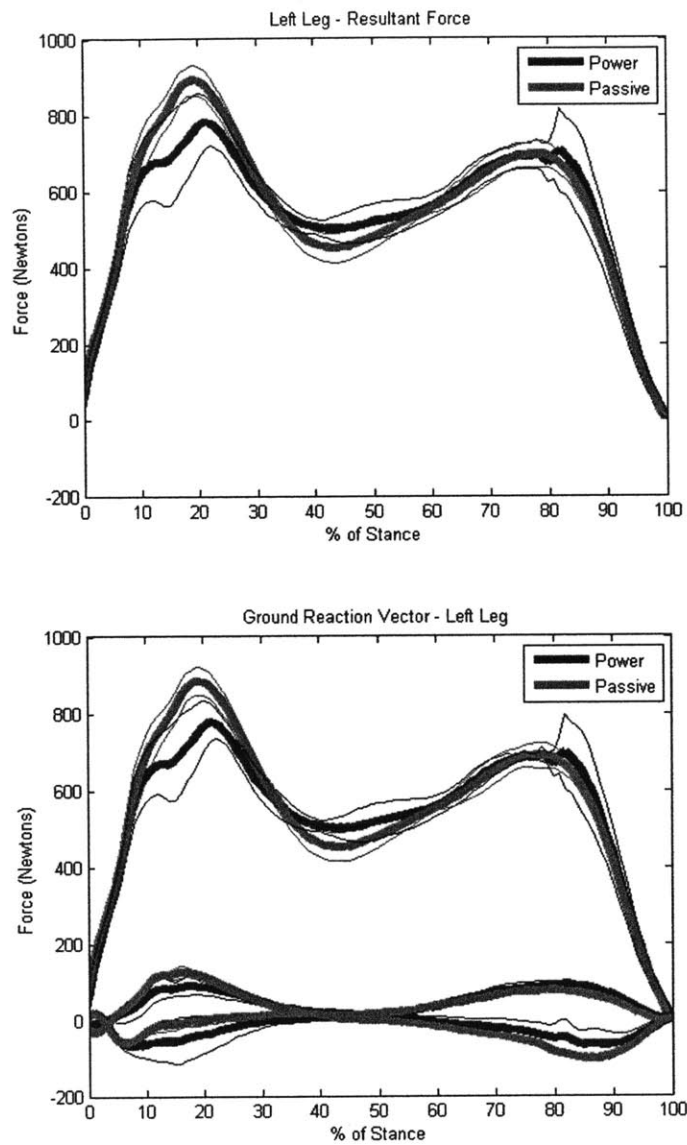


Figure 4-12: Ground Reaction & Axial Forces of Leading Leg – Subject #3 (1.25 m/s)

4.4.3.2 1.75 m/s

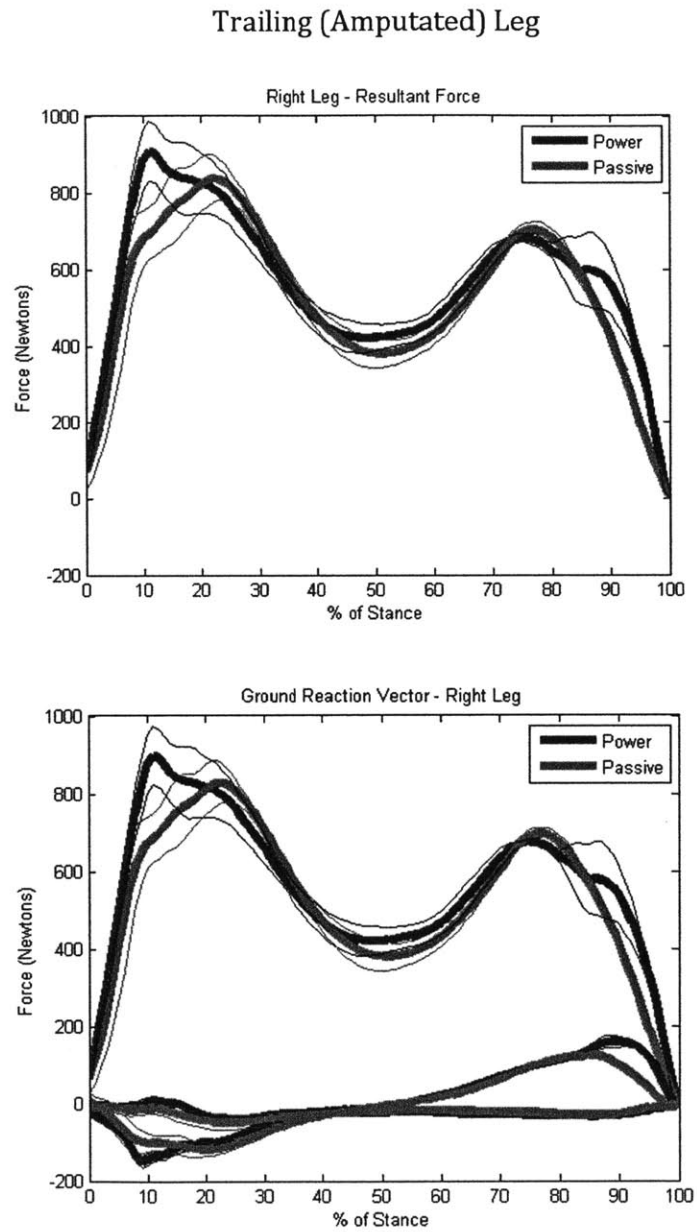


Figure 4-13: Ground Reaction & Axial Forces of Trailing Leg – Subject #3 (1.75 m/s)

Leading (Unaffected) Leg

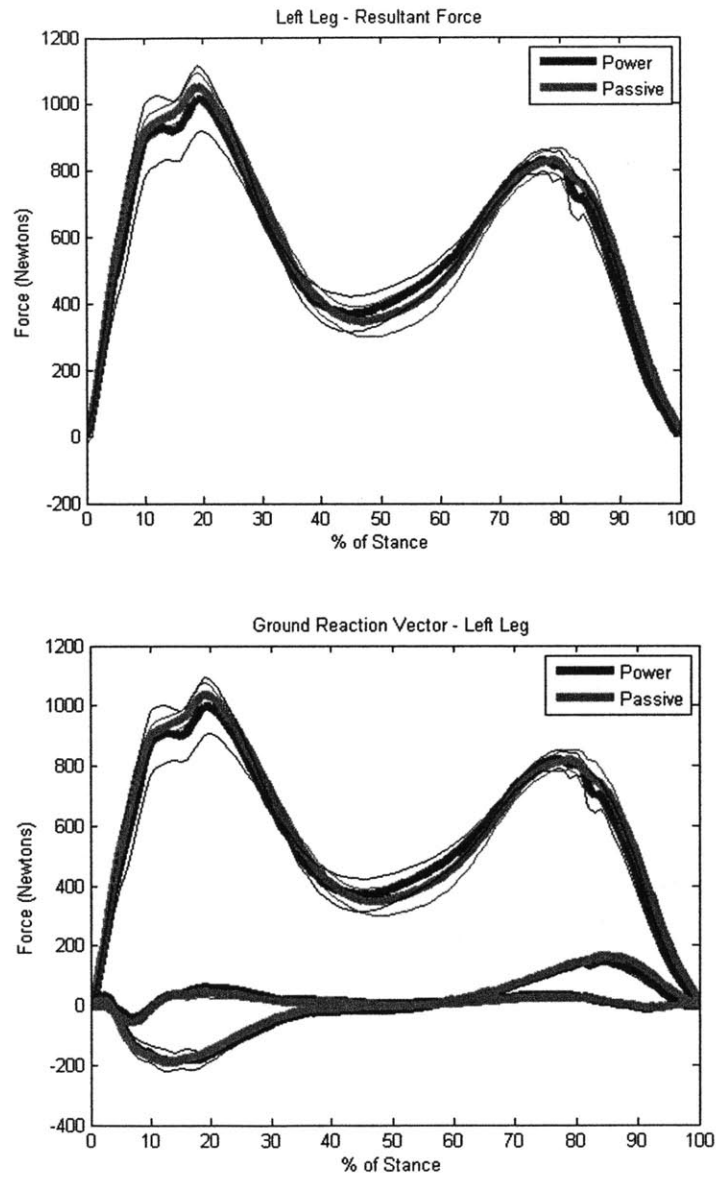


Figure 4-14: Ground Reaction & Axial Forces of Leading Leg – Subject #3 (1.75 m/s)

4.4.4 Discussion

Ground reaction forces allow us to derive the axial force vector that runs along each leg. From this force, we can analyze loading response to draw information on shock absorption at impact.

First, let us discuss the trailing or affected leg. A comparison of the resultant force plots for each subject show the effect that the powered ankle has on the user. There are extra peaks in the power force trajectory that appear clearly near toe-off that show when the ankle motors engage for powered plantarflexion. Additionally, Subjects #1 and #3, the smaller subjects, appear to be drawn to the ground at heel-strike more quickly and harder when using the powered ankle due to its extra weight. It would be interesting to see how this affects gait over the course of longer trials. This emphasizes the importance of getting the weight of the prosthesis down even more. Subject #2's power force trajectory, however, follows a path very similar to his passive force trajectory.

Now, let's discuss the axial forces of the leading or unaffected limb, specifically focusing on loading response only. Each trial for each subject, with the exception of low speeds for subject #2, show at least a slight decrease in impact force magnitude and timing. Usually, the trajectories follow the same trajectory for a while until the power trajectory falls slightly below the passive trajectory. The percentage differences for each subject are approximately 6.7%, -6.7%, and 12.2%, respectively at 1.25 m/s and about 4.8%, 6.2%, and 5.4%, respectively at 1.75 m/s. This implies that power plantarflexion of the PowerFoot does effect shock absorption, reducing the amount of stress applied to the contralateral limb at impact. This is key for improving the long-term health of the leg opposite amputation. Now that it isn't forced to compensate as much, there should be less discomfort on this side.

4.5 Pressure Characteristics

We analyzed pressure distribution at two different regions: the socket and the foot. Results from both regions are shown in the following section.

4.5.1 Socket Pressure

Measuring socket pressure, reliably, presented us with many more problems than we originally anticipated. It turns out that our sensors degraded quite a bit over the course of our walking trials. The following are example pressure readings taken from the sockets of each subject.

Subject #1

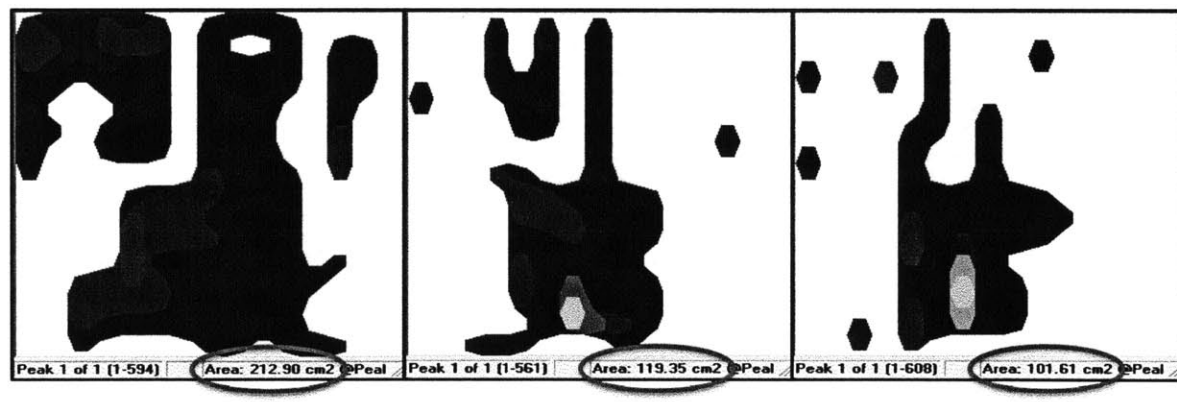


Figure 4-15: Socket Pressure Mappings – Subject #1
<First trial on left – Last trial on right>

Subject #2

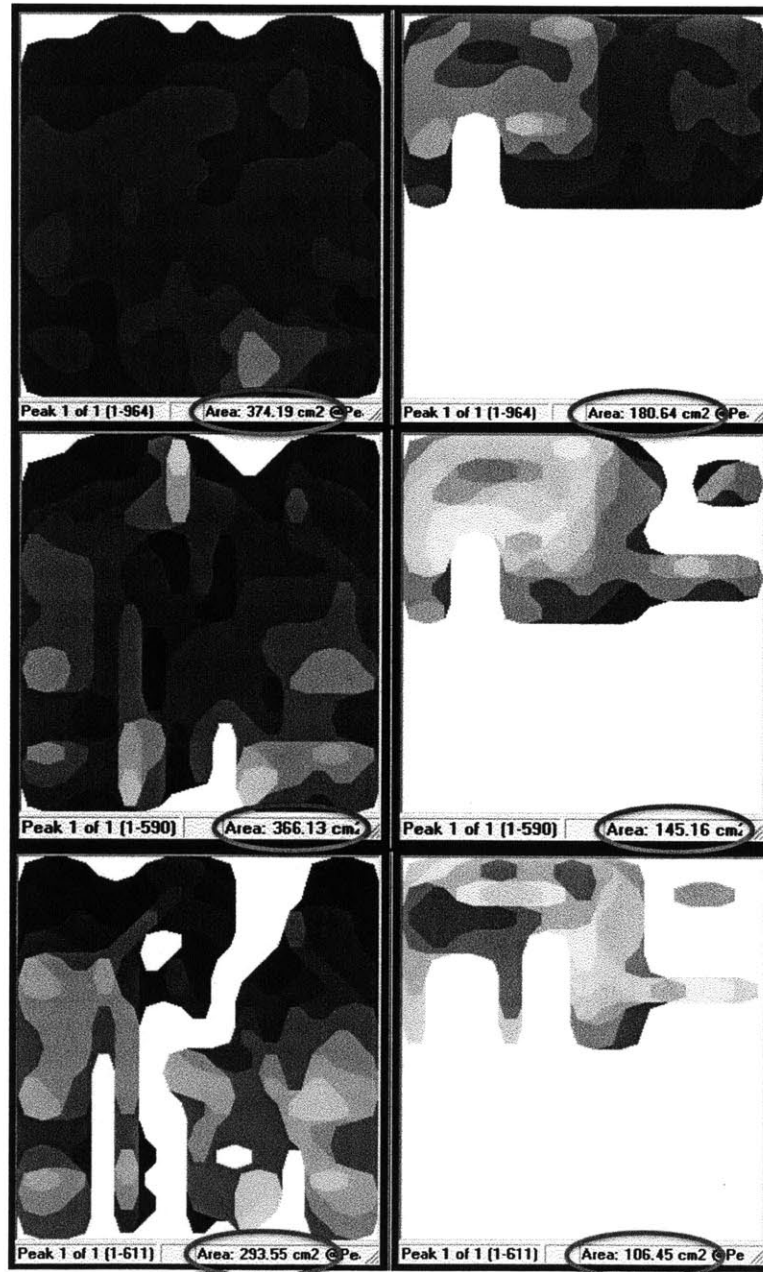


Figure 4-16: Socket Pressure Mappings – Subject #2
<First trial on top – Last trial on bottom>

Subject #3

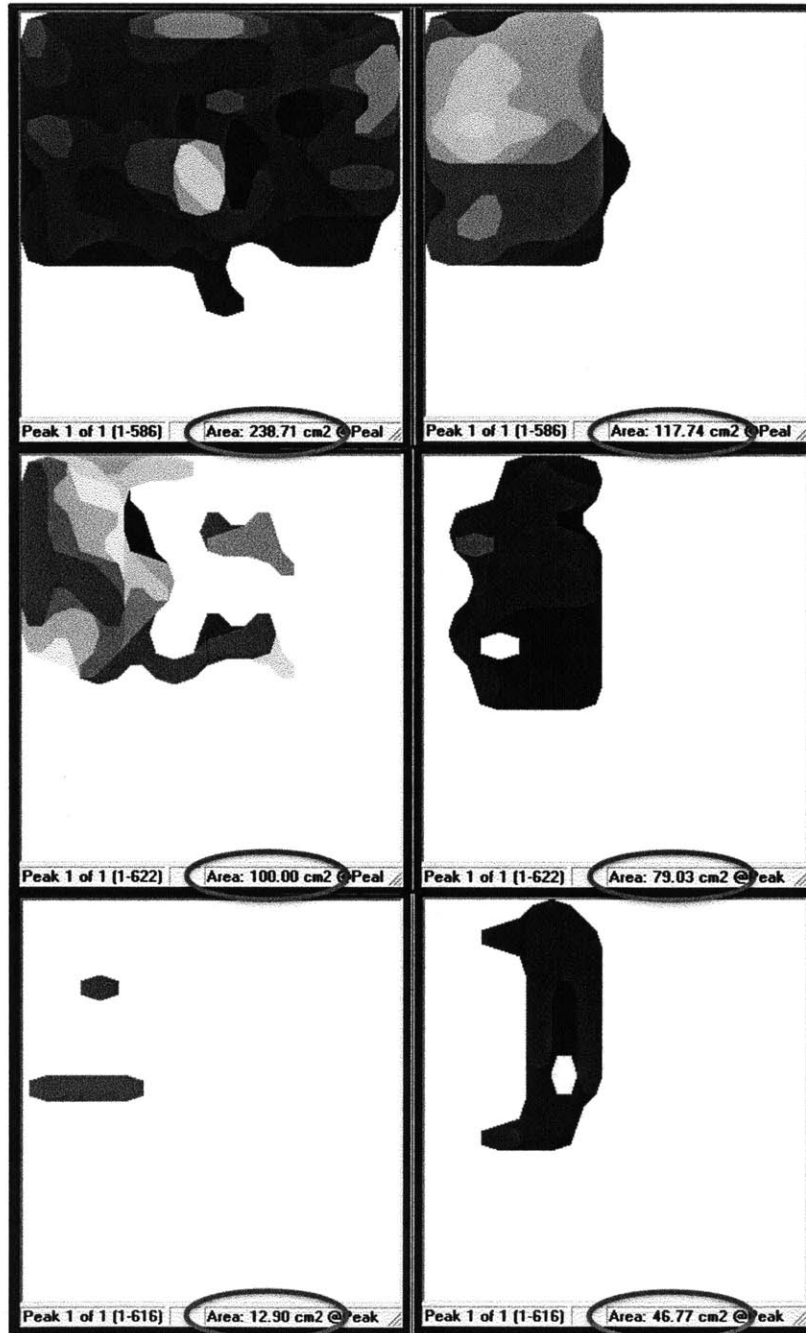


Figure 4-17: Socket Pressure Mappings – Subject #3
<First trial on top – Last trial on bottom>

4.5.1.1 Discussion

Images from Figure 4-13, Figure 4-14, and Figure 4-15, show socket pressure mappings for each subject. Each individual square window represents an individual 15 sensel by 16 sensel grid patterned sensor. To clarify, there are three separate sensor maps (one per trial) in Figure 4-13, and six separate sensors (two per trial) in Figures 4-14 and 4-15. Most subjects required two sensors to cover the entire residual limb, one on the anterior side of the residual side and one on the posterior side. Subject #1 did not.

The color-coded displays represent pressure distribution along the sensors. In these cases, each window is showing the maximum pressure recorded by each sensel throughout the course of an entire trial. Therefore, every cell activated during a trial shows up. Since all activated cells are displayed, these shots are a visual indicator of the area contacted during a trial and thus live sensing elements. Observing these figures show a pretty clear degradation in mappings going from the early recordings to the final recordings.

We can quantify the declining activity of each sensor by paying attention to the associated contact areas displayed at the bottom of every window (circled in red for every sensor). The sensors that fared the best belonged to Subject #2. We see that his only experienced an area decrease of about 28% falling from $\sim 555 \text{ cm}^2$ to $\sim 400 \text{ cm}^2$. The other two subjects experienced much larger declines, with Subject #3 having the greatest. The sensors used for Subject #3 lost 83% of their active cells over the course of the study.

Failure of the sensors is primarily due to two main factors. The sensors are incredibly vulnerable to sheer, which is huge along the vertical socket walls. Additionally, cutting the sensors contributes to their vulnerability, and we had to cut them in order for them to fit properly into the socket.

This degree of degradation in our socket sensors renders them effectively useless in this study. In the future, we must explore different ways of sensing pressure inside of the

prosthetic socket that yield accurate measurements and are durable enough to withstand our trials.

4.5.2 Foot Pressure

We were able to collect pressure from inside of the shoe of the unaffected limb. Here, we will show and discuss our findings.

4.5.2.1 Center of Force (COF)

In addition to basic foot pressure, we will also plot center of force. Center of force is another great indicator of how pressure is distributed along the foot. It also represents range of motion. COF plots will look like simple rectangular graphs with a curved line lying vertically. The vertical lines represent the displacement of the center of mass. One can imagine the displacement lying along a foot, as seen in Figure 4-16.

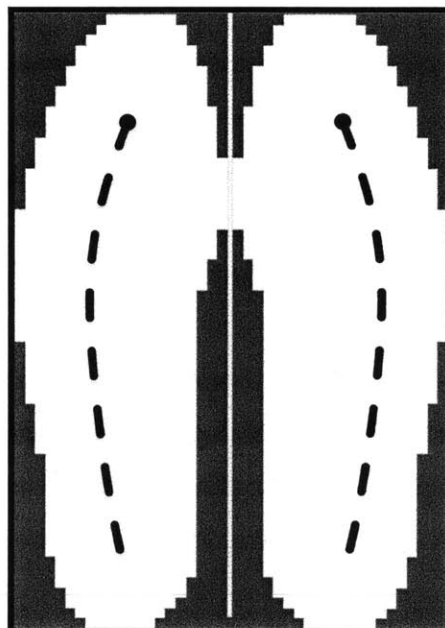


Figure 4-18: Foot Center of Force

4.5.2.2 Subject #1

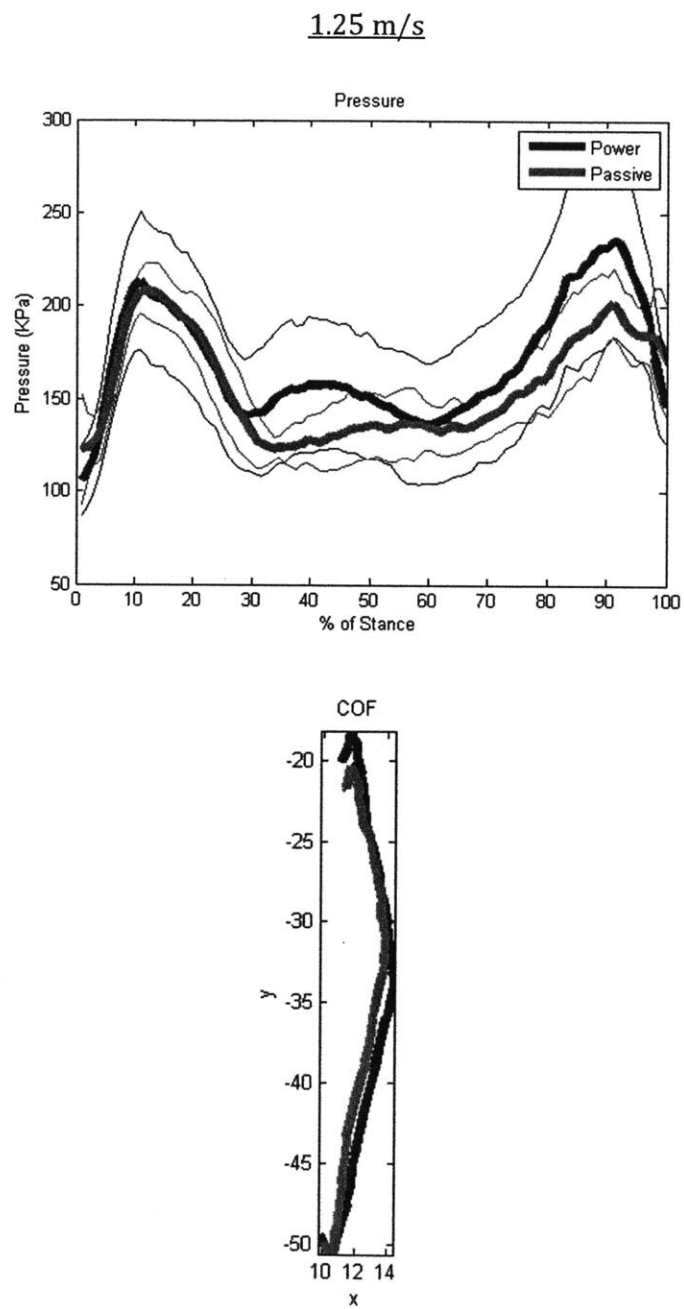


Figure 4-19: Contact Pressure & COF – Subject #1 (1.25 m/s)

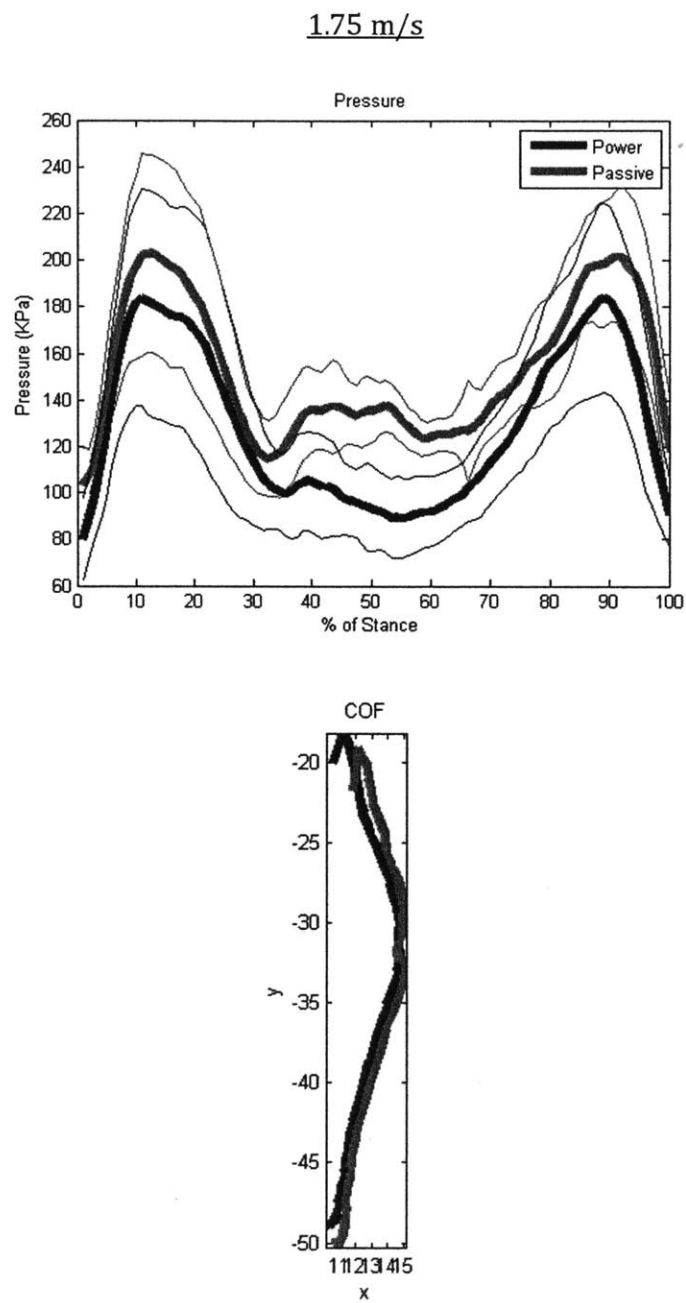


Figure 4-20: Contact Pressure & COF – Subject #1 (1.75 m/s)

4.5.2.3 Subject #2

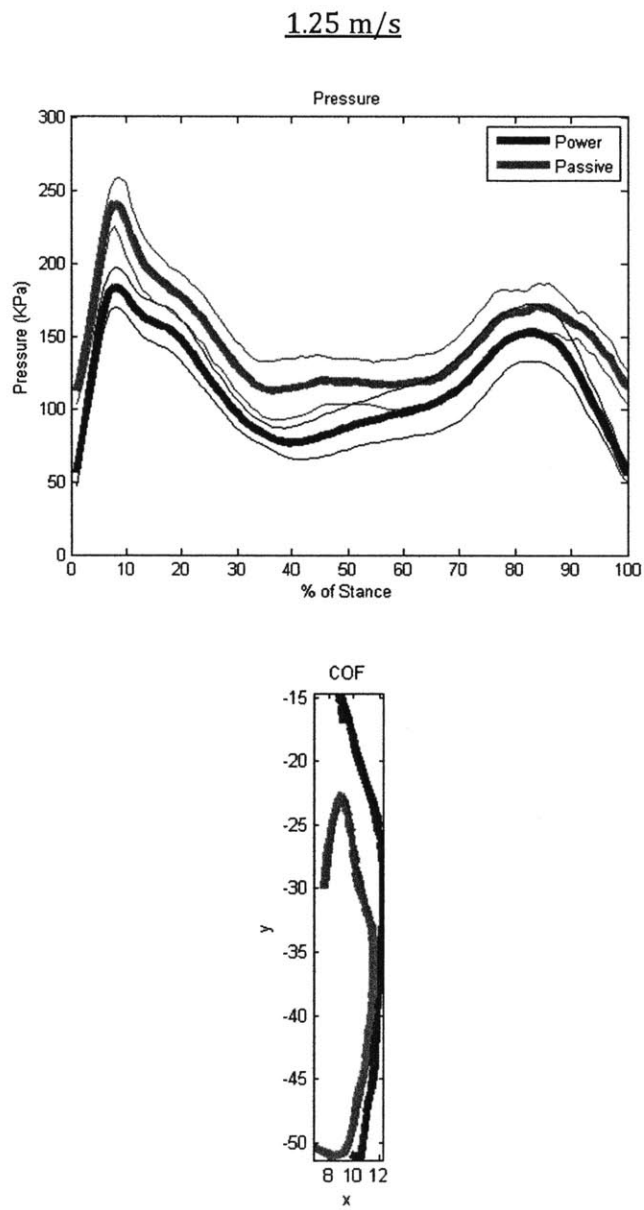


Figure 4-21: Contact Pressure & COF – Subject #2 (1.25 m/s)

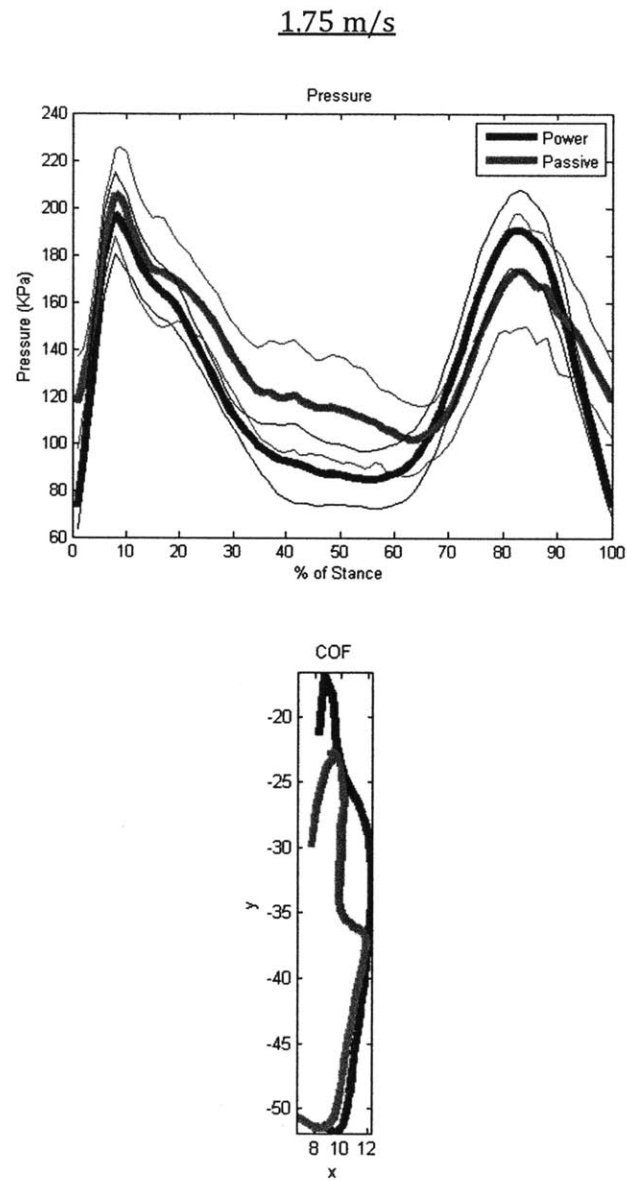


Figure 4-22: Contact Pressure & COF – Subject #2 (1.75 m/s)

4.5.2.4 Discussion

Foot pressure analysis uncovers a few very interesting things. Keep in mind that the data presented here comes from the biological foot (unamputated foot). This, yet again, allows us to study how a change in prosthesis may affect the behavior of the limb opposite amputation. Since feet are vulnerable to skin defects resulting from excessive pressure, we should see what changes in the prosthesis make any positive difference in the distribution of forces along the foot.

We will start with the center of force plots for both subjects. Both displayed at least slight differences when walking with the PowerFoot instead of their passive prosthesis. Subject #1 primarily showed similar COF trajectories, except for slight differences in heel-strike and toe-off points. His passive trajectory at the lower speed was slightly shorter than the powered trajectory. However, the effects on subject #2 were quite large. He showed a much longer, more natural looking trajectory, which implies longer stride lengths. This means that the forces along the foot will be applied to a larger surface area.

Pressure, on the other hand, is a little more difficult to interpret. The plots, in general, seem to show lower pressure when using the PowerFoot more often than not. Subject #1 does exhibit an increase in power pressure at lower speeds. Just as before with the transition cost, this could be due to Subject #1 being below the recommended size range for the powered prosthesis. The weight of the device could cause more force, and therefore pressure, to be applied to the unaffected limb. This is implied by the fact that the power pressure does not substantially exceed the passive pressure until the affected limb goes into swing phase. At this point, the weight of the prosthesis becomes much more evident.

Chapter 5

Conclusion & Future Work

We have shown that foot pressure, transition cost, and axial force at impact can all be effectively reduced by use of a bionic prosthesis. This lets us know that even with the added weight of the bionic prosthesis, it's more biomimetic capabilities (i.e. stiffness, biological power output, and realistic plantar- and dorsi- flexion) are key to restoring "normal" walking patterns to amputees. Each of these parameters is a key contributor to common health issues faced by amputees. The improvements shown here are pretty significant because a reduction of stress applied to the average amputees' legs over time will go a long way toward minimizing many of the long-term complications they face.

In the future, we will continue to see these types of prosthetic devices develop to become lighter, smarter, and more powerful. The weight issue in particular is evident in our results, with the smaller subjects' benefits being limited by the added weight of the powered prosthesis. As these improvements come to pass, amputees will inch closer and closer to non-amputee gait. They will also be closer to living a life free of complaints (at least complaints due to the prosthesis). This work underscores the importance of developing more biomimetic prostheses and assistive devices.

5.1 Contributions

This thesis was an investigation of the factors that influence excessive loads applied to the lower extremities of transtibial amputees. We have performed a thorough analysis of axial forces, pressure distribution, and shock absorption to uncover the differences produced by using prostheses with more biomimetic capabilities. This study contributes to the science of rehabilitation and human-machine interfacing. We hoped to underscore the importance of building assistive medical devices that are biologically inspired and intelligent. We have laid the groundwork for the development of new ways of monitoring pain and discomfort when forming device attachments to the body.

5.2 Future Work

Human subject testing for this project was limited by time and hardware. Trials were conducted over the course of five hours, with equipment setup, conventional foot testing, PowerFoot fitting, PowerFoot tuning, and PowerFoot testing, all being done within this timeframe. This means that subjects generally had about a 30 minute to one hour training period on the PowerFoot before we collected data. It would be nice to see the data changes when subjects are given a much longer training period (a few weeks or so) with the PowerFoot. In the future, we could supply each subject with his own PowerFoot to tryout for multiple weeks, and then conduct the study after the user has effectively adapted to the new prosthesis. Unfortunately, we did not have the time or equipment to conduct this type of study this go around.

This study has several potential applications, both scientific and engineering. Perhaps the most powerful applications will be in the development of future assistive devices. This thesis will aid with the attachment problem and contribute to new prosthetic control schemes. One of the most daunting challenges in human-machine interfacing is the attachment problem, which basically means figuring out the best ways to attach devices to humans in ways that augment and not hinder them. The comfort assessment proposed in this thesis will help govern new methods of attachment that do not harm the wearer.

Additionally, this project will contribute to innovative control strategies that optimize for comfort. Smart devices offer extended freedom in tuning. We can now use this freedom to adjust parameters such that we minimize discomfort.

Each of these applications are a few years down the road, but with the proper work, they could forever change the way we think about augmenting humans and connecting to the body. As devices continue to develop, we will have more data monitoring capabilities. Now we have a foundation for how to use this data for optimizing comfort.

Appendix A

Tekscan/Vicon Synchronization

Synchronizing the Tekscan F-Scan system with the VICON Motion Capture system calls for the use of a dedicated GPO file that defines an output synch signal generated by VICON. This signal will be used to communicate a start and stop event to Tekscan. The GPO file is a simple xml script written to define the type, polarity, start/stop events, and duration of the synch pulse. A copy of the .gpo file is shown below:

```
<?xml version="1.0" standalone="yes"?>
<AllPrograms>
  <Program Name="Reverse_duration">
    <Type>Duration</Type>
    <Polarity>Low</Polarity>
    <StartEvent>StartCapture</StartEvent>
    <StopEvent>StopCapture</StopEvent>
    <StartOffset Frames="" MicroSeconds=""/>
    <StopOffset Frames="0" MicroSeconds="0"/>
    <PulseWidth Frames="0" MicroSeconds="0"/>
    <PulsePeriod Frames="0" MicroSeconds="0" Ticks="0"/>
  </Program>
</AllPrograms>
```

This script is to be loaded into the VICON Nexus software. Once loaded, Tekscan can be setup to accept commands from this signal to start and stop recordings. Finally, the two devices are synchronized.

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